ENHANCING LOCOMOTOR PERFORMANCE BY MODULATING SHOE CUSHIONING PROPERTIES

By

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To my parents, who always pushed me to accomplish more than I ever thought I could

and

To Sarah

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CHAPTER 1

INTRODUCTION AND REVIEW OF LITERATURE

1.1 Motivation and Purpose

The overarching goal of this work is to determine if and how footwear properties can be modified to alter biomechanical aspects of movement. There are many useful applications for this knowledge, from designing a running shoe that that improves running economy to designing footwear for gait rehabilitation or symmetry. This work will investigate the biomechanical effects of sole cushioning properties, specifically hardness, on walking kinetics.

Technology is advancing to the point where shoe midsoles can be custom-designed to meet individual cushioning needs or preferences; for example, through Selective Laser Sintering (SLS) of powdered elastomeric plastics (New Balance 2015). In order to make the most of this new technology for all people, runners and walkers alike, the body of knowledge concerning the relationship between midsole cushioning parameters and locomotor performance needs to be understood more thoroughly. The goal of this thesis was to, first, create a testing shoe with the capacity to have its cushioning properties varied mid-experiment, and second, have participants walk on various cushioning conditions within one experimental session while measuring biomechanical performance. We sought to quantify how critical walking outcome metrics, defined later in this section, change as a function of midsole cushioning parameters, in order to answer the overarching question: can we manipulate shoe properties to enhance aspects of human walking? A secondary question of interest was then to assess: and are these enhancements subject-specific or ubiquitous (i.e., consistent across individuals)?

To evaluate these questions we focused on two key outcome metrics related to human walking performance: push-off and collision kinetics. Walking models and experimental evidence (Donelan et al. 2002a, Kuo et al. 2005, Kuo 2002) indicate that the two phases of walking gait that exhibit the largest center-of-mass (COM) work values are push-off (at the end of stance phase immediately before leg swing) and collision (just after foot contact). Quantifying the mechanical work or power performed during push-off and collision are common ways to assess if changing conditions (e.g., the properties of the contact surfaces at the heel and forefoot during walking) has notable effects on gait (see Section 1.4.4).

The cost of cushioning hypothesis (Frederick et al. 1983) postulates that the body must expend energy to cushion the body from running impacts when the collision interface (heel, midsole and/or ground surface) does not adequately attenuate the impact forces. Empirical evidence has been observed that supports this claim (Frederick et al. 1983, Tung et al. 2014, Kerdok et al. 2002, Hardin et al. 2004, Boyer and Nigg 2004, Nigg et al. 2003) (see Section 1.4.3). Since the characteristics of ground impact can have a significant effect on movement biomechanics, it is of interest to ask if collision effects during walking can be similarly altered by manipulating the cushioning under the heel. Push-off is the main phase of positive power generation, and is believed to play an important role in safe and economical gait, since push-off from the trailing leg helps to reduce collisions and energy loss during leading leg footstrike collisions (Kuo 2002). It is therefore of interest to evaluate if and how push-off characteristics can be tuned by altering footwear cushioning at the forefoot.

In this thesis, I sought to investigate a large number of heel and forefoot cushioning combinations in order to perform a systematic and fairly comprehensive investigation of cushioning properties. I used biomechanical outcome measures related to push-off and collision to gain insight into how cushioning affects gait biomechanics. If biomechanical trends in COM peak collision and push-off power can be determined as cushioning is altered, and especially if the effects of the forefoot and heel cushioning can be decoupled, it could be possible to tailor footwear properties for different purposes, either on a subject-to-subject basis or more ubiquitously.

1.2 Metabolic Measurements and Their Importance

A common metric used to understand locomotion is metabolic energy expenditure, which provides insight on the economy of movement. Footwear or other devices that reduce metabolic expenditure (increase economy) can be exploited to allow for a multitude of benefits. They may allow runners to improve their race times or to increase their maximum distance traveled, or walkers to locomote with less demand on their muscles.

There are two common methodologies for measuring metabolic expenditure: direct and indirect calorimetry (Dauncey 1980). The first (direct) takes advantage of the predictable and measurable heat produced as a byproduct of the metabolism of glucose into usable biological energy (Burton 1935), and the second primarily uses oxygen uptake (and possibly carbon dioxide production and nitrogen excretion through urine) as the key measurement (Ferrannini 1988). Due primarily to the ease of experimental design, indirect calorimetry is used most often in the field today.

Though indirect calorimetry provides a method of expenditure estimation, its use also has critical drawbacks. Firstly, data from locomotion trials must be recorded at “steady state”. This means that as conditions vary between trials, the locomotion must be maintained until the body of the subject is aerobically metabolizing at the adequate amount for the activity; typically trials must last for greater than six minutes (Dauncey 1980). For this reason, trials tend to take upwards of eight minutes each, with only the last few minutes being viable for calorimetric measurements (Griffin et al. 2003). Secondly, across multiple experiment days, there is intra-subject variability of metabolic expenditure estimates based on factors like recent food intake and time since the last meal (Dauncey 1980), since the amount of energy stored locally in and around muscle cells alters the precise metabolic oxygen needs of the participant. For these reasons, though metabolic analysis plays an important role in biomechanics and understanding the effects of many variables on the metabolic cost of multiple activities, there are often other biomechanical metrics that can quantify movement in other interesting ways, especially when performing large parameter sweeps with this outcome measure.

1.3 Mechanical Measurements and Their Importance

COM work rate and COM power provide a different perspective on locomotion. In walking, COM power is measured using a force-platform (Cavagna 1975) mounted either on the ground or on a split-belt treadmill. From here, forces and mass measurements can be used to obtain acceleration of the body’s COM, and integrated to yield velocities (Cavagna 1975, Donelan et al. 2002a), which can be used with the forces to obtain power curves during locomotion. It is common to perform this calculation using data from separate force plates for each leg during walking, since the double-support phase of walking involves both the generation and dissipation of mechanical energy from the trailing and leading limbs (Donelan et al. 2002a). When total positive COM power (i.e., power that biomechanically must have been produced by energetic muscle actions or elastic return) is summed over a stride and divided by stride period, the resulting outcome provides insight into the mechanics of walking. Under certain conditions, the trends found in COM power when conditions are varied tend to correlate with trends in metabolic cost (see Figure 1), though the relationship between biomechanical measures (e.g., work, power) and metabolic measures is highly non-linear and not yet fully understood (i.e., at the whole-body level we have no formula that directly maps between these two metrics).

Griffin et al. (2003) performed an experiment that measured how adding significant mass to walking subjects affected walking metabolic energy expenditure and COM power, and found that similar trends arose in both these metrics (Figure 1). Though the shapes of the curves displayed do not perfectly match and the magnitudes of the data points are distinctly different, if an increase in speed resulted in an increase in metabolic expenditure, it also resulted in an increase in COM power. Similarly, if an increase in mass resulted in an increase in COM power, it also resulted in an increase in metabolic expenditure.

COM work can be computed by integrating COM power over time. Integrating only over the push-off phase of walking yields push-off work, and integrating during the collision phase yields collision work. This technique has been used elsewhere in literature to characterize specific aspects of the gait cycle as it pertains to energetic phenomena (Zelik and Kuo 2010, Zelik et al. 2015). Comparing different aspects of COM power curves and COM work can reveal a lot about walking gait and the energetic differences that occur when imposed conditions, such as sole cushioning properties, are altered. Due to the ease of comparison elicited by COM power (it can be calculated from ground reaction forces, and only requires a few strides, i.e., a few seconds), it is a useful metric for determining the effects of midsole cushioning on gait.

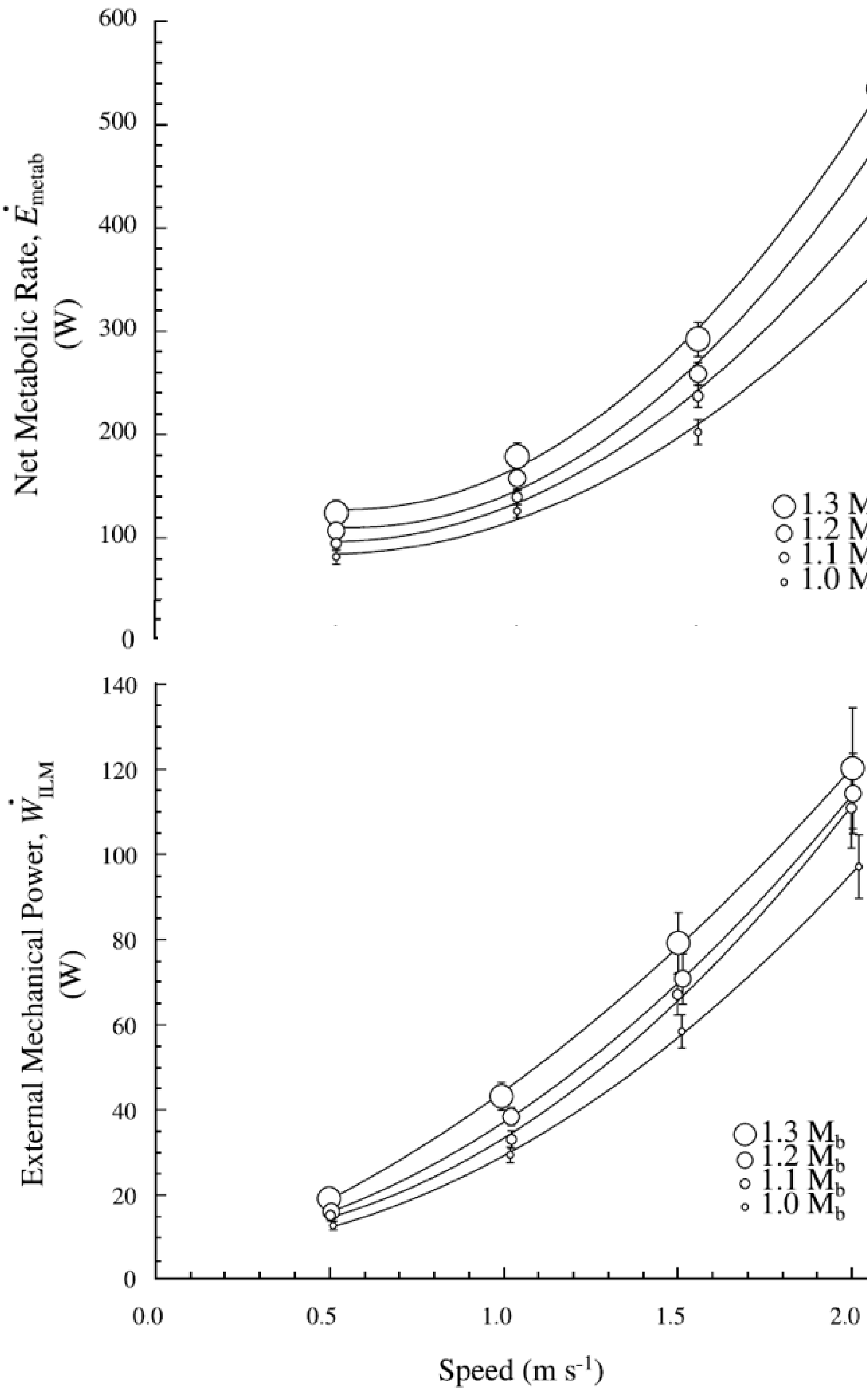


Figure 1: Effect of additional mass on the body and increasing walking speed to both metabolic expenditure and COM power (referred to here as external mechanical power). Both as speed increased and as mass increased, COM power and metabolic power showed similar comparative trends of an increase in the represented metric (adapted from Griffin et al. 2003).

Another useful metric to give insight into how sole cushioning may be affecting the mechanics of walking gait is joint powers. Using motion capture, instrumented force platforms, and a rigid-link model of the body (Cappozzo et al. 1976, Gordon et al. 1980), inverse dynamic estimates of joint power can be estimated. This technique allows an in-depth power analysis able to uncover which joints are generating and which joints are absorbing power over the gait cycle. Validations of this technique have been performed (Zelik and Kuo 2010, Zelik et al. 2015), and it has been concluded that the derived estimates for joint power do not fully match total power (defined here as COM power plus the mechanical power needed to move extremities relative to the COM) without a more rigorous (six degree-of-freedom) analysis technique (for example, the rigid-link joint technique underestimates push-off work by almost 30% compared to the six degree-of-freedom technique), and even then some unmodeled energetic aspects of gait cause error. These drawbacks related to absolute accuracy are less relevant when limiting the use of the inverse dynamics estimates to comparing relative joint power changes across sole cushioning conditions. Since the model does not change across conditions, these relative comparisons can lend insight to the underlying sources of the energetics of locomotion. When considering joint-level biomechanics, the focus of this research is therefore on assessing relative changes in power and work.

A metric that can facilitate the estimates of power dissipated in the foot-shoe complex is deformable foot power. The foot and shoe can be modeled as a deformable body and an algorithm can be employed to calculate the power (generally dissipated) by distal foot elements. This technique utilizes center-of-pressure changes and foot COM position, velocity, and angular velocity to capture work performed by soft tissue and shoe deformations, which is not modeled/captured by more traditional rigid-body inverse dynamics joint power estimates (Siegel et al. 1996, Takahashi et al. 2012, Takahashi and Stanhope 2013).

1.4 Conditions That Affect Locomotion

The information presented in this section directly informed the experimental design (see Chapter 2) as well as the posed hypotheses (see Section 1.6). To know which variables to control for in this experiment and what trends have already been presented in previous works, it was important to conduct a thorough review of literature.

1.4.1 Mass

One of the more commonly referenced relationships with regard to altering footwear properties for runners is the increasing effect of mass added to the foot on metabolic cost of locomotion. It was found that for every extra 100 grams of shoe mass, running metabolic cost increases by roughly 1% (Frederick 1984). This was measured at the time using open circuit oxygen uptake, using a formula similar to that outlined by Brockway (1987). In that work, it was concluded that a formula can be used, within a reasonable margin of error, to estimate metabolic cost using only oxygen intake as a variable, scaled only by a constant coefficient (Brockway 1987). There have been numerous studies since then that have corroborated the trend of decreased running economy with increasing mass on the foot (Martin 1985, Divert et al. 2008, Burkett 1985, Hanson et al. 2011, Franz et al. 2012, Tung et al. 2014). Figure 2 demonstrates the trend put forth by Frederick (1984) as substantiated by an experiment performed by Franz et al. (2012).

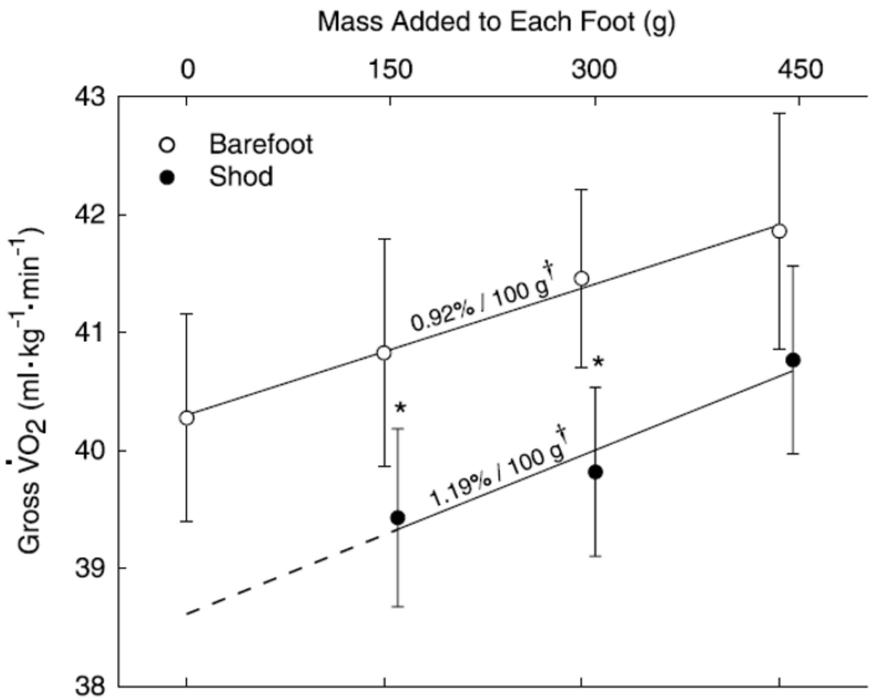


Figure 2: Effect of added mass to the foot on gross oxygen uptake. The trend towards more oxygen uptake with increasing mass on the foot can be seen at a rate similar to that predicted by Frederick (Franz et al. 2012).

Though less relevant to the focus of this thesis, it should be noted that addition of mass to other parts of the body, specifically more proximal than the foot, showed a weaker but still present trend towards increasing metabolic cost. As an example, one study (Martin 1985) performed a condition where mass was added to the thighs instead of the feet. The result of that experiment showed that the impact of increasing thigh loading on oxygen uptake was around half that of increasing foot loading (Martin 1985). Figure 3 shows this trend by comparing the change in oxygen uptake from the unloaded condition as loads are applied to the feet and to the thigh.

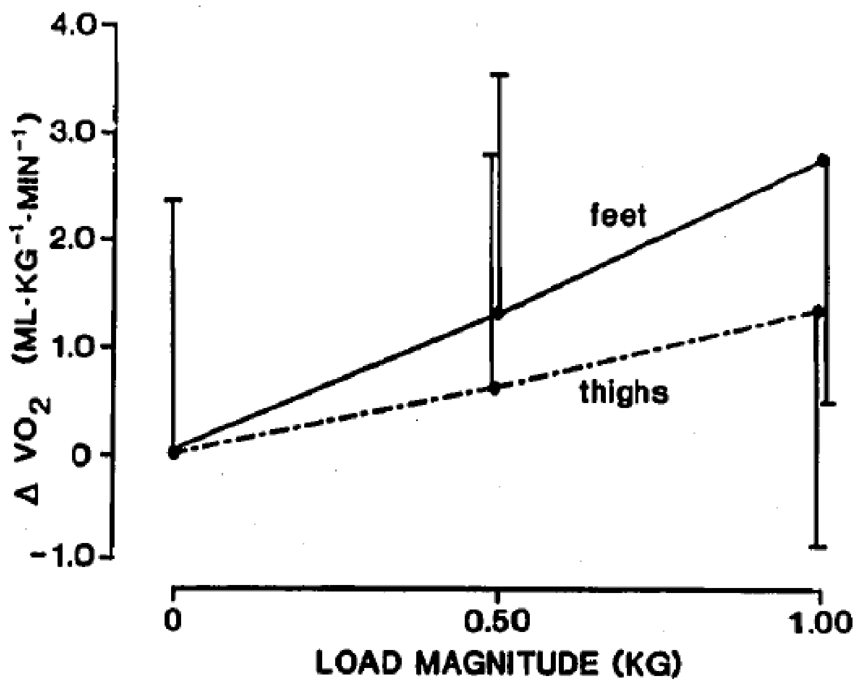


Figure 3: Effect of added mass to the foot and thigh on increase in oxygen uptake. The value for uptake increases due to the addition of mass at the thigh at roughly half the rate that it increases due to the addition of mass at the foot (Martin 1985).

Furthermore, as would be expected, adding mass to the trunk also results in an increase in the metabolic cost of walking (Griffin et al. 2003). Figure 1 shows this trend for four walking speeds and four loading conditions, ranging from 0.5 m/s to 2.0 m/s and from one body mass to 1.3 body masses. Taken together with the findings of Martin (Martin 1985), it can be concluded that adding mass specifically to the foot will likely increase the metabolic cost of walking.

Due to these reported effects of changing mass on the energetics of walking, it is important to design an experiment in such a way that footwear conditions are mass-matched. This way, comparisons between conditions can be made and the effect of the independent variable can be decoupled from a potential resulting change in mass.

1.4.2 Velocity

Both metabolic and COM power estimates increase with walking and running speed (Griffin et al. 2003, Margaria et al. 1963, Zelik and Kuo, 2010). Figure 1 (seen in Section 1.3), beyond showing just the mass dependence of metabolic expenditure and COM power during walking, also shows its speed dependence (Griffin et al. 2003). In another study (Kuo et al. 2005), the experimenters also discovered that averaged positive COM power increased with walking speed.

One study (Margaria et al. 1963) characterized the gait velocity-metabolic expenditure relationship clearly by using a treadmill and oxygen consumption measurements at steady-state. As can be seen in Figure 4, both walking and running show an increasing relationship between speed and metabolic expenditure for level locomotion, and locomotion up and down 5% grades. Running has the added interesting quality of seemingly showing a linear relationship between speed and metabolic cost, with the extrapolated origin near basal metabolic expenditure (Margaria et al. 1963).

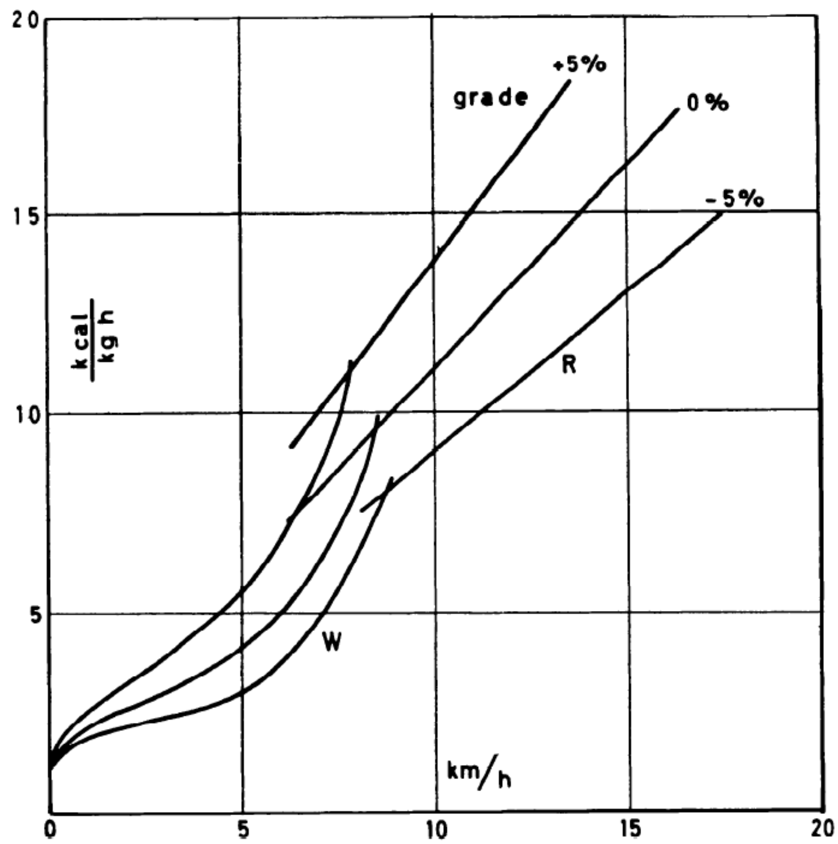


Figure 4: Effect of increasing locomotion speed on metabolic expenditure. 'W' in the Figure represents walking curves at the shown grades while 'R' represents running curves. Note the linearity of the running curves (Margaria et al. 1963).

Velocity also affects other biomechanical aspects of gait. One study (Zelik and Kuo, 2010) related joint power calculations from inverse dynamics (see Section 1.3) and COM power at various walking speeds. First, it was found that the ankle tended to be more plantar-flexed at push-off, the knee tended to be more flexed at heel-strike, and the hip tended to be more extended at heel-strike as speed increased. Second, it was found that all the joints showed larger magnitude power peaks at various points in the gait cycle. This would be expected since both metabolic and COM power increase with walking speed (Griffin et al. 2003). Further results of this experiment will be discussed later.

During running, it has been observed (Arampatzis et al. 1999) that as gait velocity increases on a given surface, the stiffness of the equivalent spring meant to model the human leg, estimated using the spring-loaded inverted pendulum (SLIP) model of running, tends to increase. Though this has been a point of dispute in the past (McMahon and Cheng 1990), this correlation seems to be more supported experimentally. This relationship between leg stiffness and velocity implies that the leg (through the knee and hip especially) tends to go through a larger range of extension and flexion as running speed increases (Arampatzis et al. 1999).

Since changing velocity tends to alter the kinetics and kinematics of walking, in order to make measurements which demonstrate only the effect of the independent variable, velocity must be controlled between conditions as well as between subjects.

1.4.3 Surface and Shoe Stiffness (Running)

Understanding how the body alters running gait due to changing properties of the contact surfaces (i.e., foot-sole-ground) can also inform predictions as to how the body will react to changing surfaces during walking.

The physical characteristic of a given running surface have been determined to play a role in how one runs. One study (Kerdok et al. 2002) showed that increasing surface stiffness in the range of 75.4 to 945.7 kN/m has a significant effect on running biomechanics. Of note, the metabolic expenditure as calculated from oxygen uptake measurements showed a positive trend with increasing surface stiffness ($P=0.0001$). The experimenters did not examine effects of altering midsole properties, but did control for potential effects by ensuring that all runners wore the same shoes during the experiment. The running speed over the entire experiment was 3.7

m/s. Figure 5 shows the increasing metabolic power as a function of surface stiffness. Note the logarithmic scale on the surface stiffness axis.

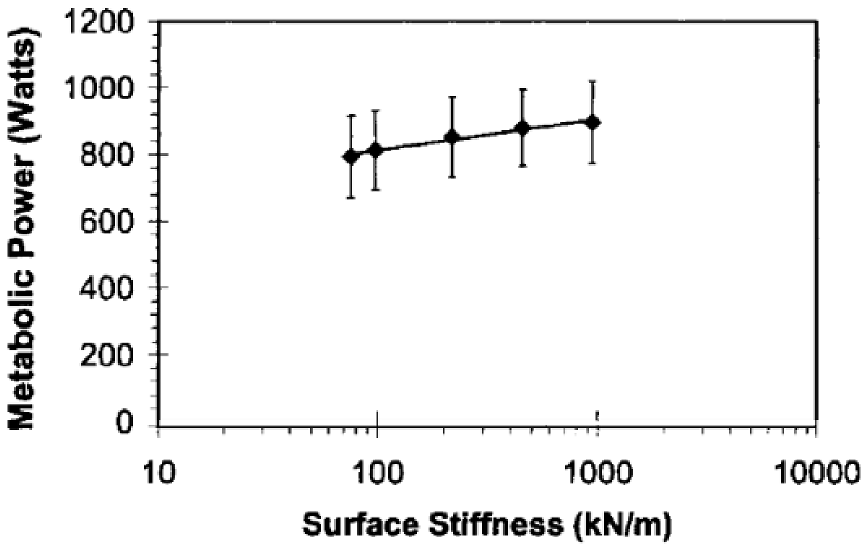


Figure 5: Effect of surface stiffness on derived metabolic power. As the surface stiffness of the treadmill was increased, the metabolic power expended by the runners also increased. Note the logarithmic scale of the surface stiffness axis (adapted from Kerdok et al. 2002).

Evidence suggests that runners tune their leg stiffness based on surface properties in order to maintain roughly constant support mechanics (Ferris et al. 1998, Ferris et al. 1999, Kerdok et al. 2002). In this context, leg stiffness is defined as the peak force on the leg over the maximum change in distance from the hip joint to the ankle joint during stance phase. This in effect means that a runner's COM trajectory is invariant to surface stiffness due to the active tuning of their leg stiffness. Damping properties of the treadmill platform were relatively low (damping ratio reported as less than 0.1), so the platform acted primarily as an elastic member, returning much of the stored strain energy back to the runner (Kerdok et al. 2002).

Using only running surface stiffness as the independent variable in a metabolic analysis without further surface specifications can be misleading, since energy dissipation in the surface can play an important role. A later study (Hardin et al. 2004) showed that increasing surface stiffness in the range of 100 to 300 kN/m significantly ($P < 0.0033$) decreased oxygen consumption. The authors addressed the discrepancy between these findings and the findings of the previously discussed study (Kerdok et al. 2002), saying that their (Hardin et al. 2004) method of constructing a compliant surface may have introduced increased damping or inertial effects of the treadmill platform itself, possibly negating the elastic rebound effects observed in the other study. Highlighting this claim, participants of the study mentioned that running on the low stiffness conditions resembled “running on sand” (Hardin et al. 2004).

The effects of midsole cushioning properties and running surface properties may be superimposable. It was proposed by Hardin et al. (2004) that sole cushioning properties could potentially have a more pronounced effect on running metabolic expenditure when running on a harder surface. In other words, the authors conclude that in their study the effects of the treadmill cushioning overshadowed the effects of the midsoles.

Surface impact concerns may also be a determinant in the metabolic cost of running. One study (Tung et al. 2014) aimed to test the cost of cushioning hypothesis (Frederick et al. 1983). The experimental setup involved one shod condition and three unshod conditions: one on a rigid treadmill platform, one on 10 mm thickness foam slats adhered to the treadmill belt, and one on 20 mm thickness foam slats. The foam slats were the same material as the foam used in the construction of the shoe midsole. It was found that running on the 10 mm thickness foam slats resulted in an average of 1.63% lower metabolic expenditure than running on the rigid treadmill platform. However, if the foam was too thick then metabolic benefits were not observed. For instance, there was no significant decrease in metabolic power due to the 20 mm foam condition compared to the rigid treadmill platform condition. The experimenters also noted substantial individual variation across subjects, as can be seen in Figure 6. The cost of cushioning hypotheses (Frederick et al. 1983) does not disagree with the previously proposed concept of leg tuning (Kerdok et al. 2002). It instead adds another level of complexity to it, focusing not on COM mechanics and total COM vertical displacement, but rather specifically on foot-strike and impact tuning.

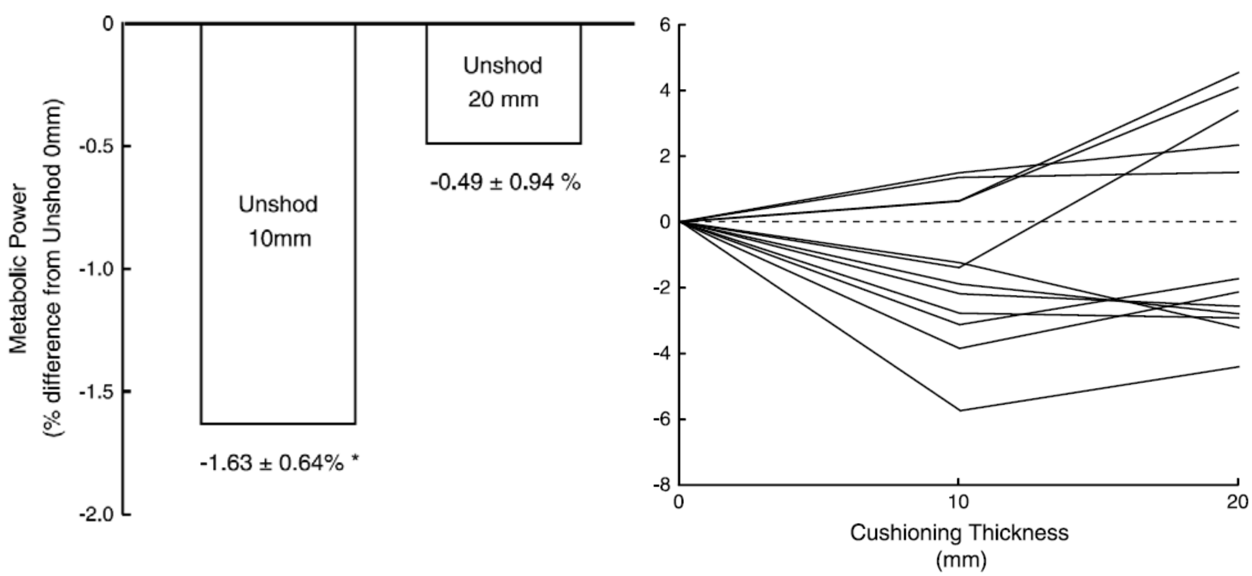


Figure 6: Effect of cushioning thickness on metabolic power of running for multiple subjects. Though the average (left) shows a significant (P = 0.034) drop in metabolic power for the 10 mm thickness cushioning condition, the subjects for the experiment all had substantial variation (right) in their adaptation to the foam cushioning (adapted from Tung et al. 2014).

Tung et al. (2014) successfully separated the effects of shoe mass and the effects of cushioning by applying soling material directly to the treadmill belts and asking the subjects to run barefoot. The results indicated that there was indeed a cost associated with cushioning the body from impacts that can be mediated by properly tuning contact surface (midsole and surface) cushioning. Due to the superposition of increased mass effects (see Section 1.4.1) on the reduced cost of cushioning, they also found that metabolic cost of running in cushioned shoes was not significantly different from running barefoot on a rigid treadmill platform (Tung et al. 2014). However, one limitation to this study was that because the same cushioning material was spread across the entire ground surface, the investigators were unable to decouple landing (collision) cushioning effects vs. push-off cushioning effects; and it may be that ideal cushioning would be different for these functionally disparate parts of the stride cycle.

Since muscles must use metabolic energy whenever they activate, the body's adaption to running surface through muscle pre-activation may be a factor in the overall metabolic demand of running. Muscles in the leg have been shown to pre-activate to tune the impact force characteristics on the lower extremities during locomotion (Boyer and Nigg 2004). The goal of this tuning may be to try to create a critically damped system to avoid soft tissue vibration (Nigg 1997). It was found that though there was no significant change in soft-tissue peak acceleration when altering shoe cushioning properties, muscle activity in the quadriceps, measured through electromyography (EMG) before foot contact, was significantly ($P < 0.05$) correlated with loading-rate, which in turn was inversely related to midsole cushioning (Boyer and Nigg 2004). This implies that cushioning more suited to attenuate impact effects may mitigate the need for muscle pre-activation, reducing metabolic expenditure.

Individual variation has been observed for the effect of midsole properties on running metabolic expenditure. Nigg et al. (2003) performed a study where two shoes of nearly identical mass (293 g and 288 g) but differing elasticity and viscoelasticity were worn during a series of metabolic tests. Though the drawbacks of metabolic cost measurements do play an important role, the experimenters tried to eliminate some variability in the measurements through clever experimental design. They eliminated order effects by designing two sets of trials, occurring during two experimental sessions, reversing order between the viscous and elastic conditions for each trial set and also for each session. The experimenters' hypotheses were confirmed when results showed no significant population trends between the viscous midsole and the elastic midsole but significant subject-specific metabolic trends (P = 0.05). Figure 7 (Nigg et al. 2003) illustrates these subject-specific trends.

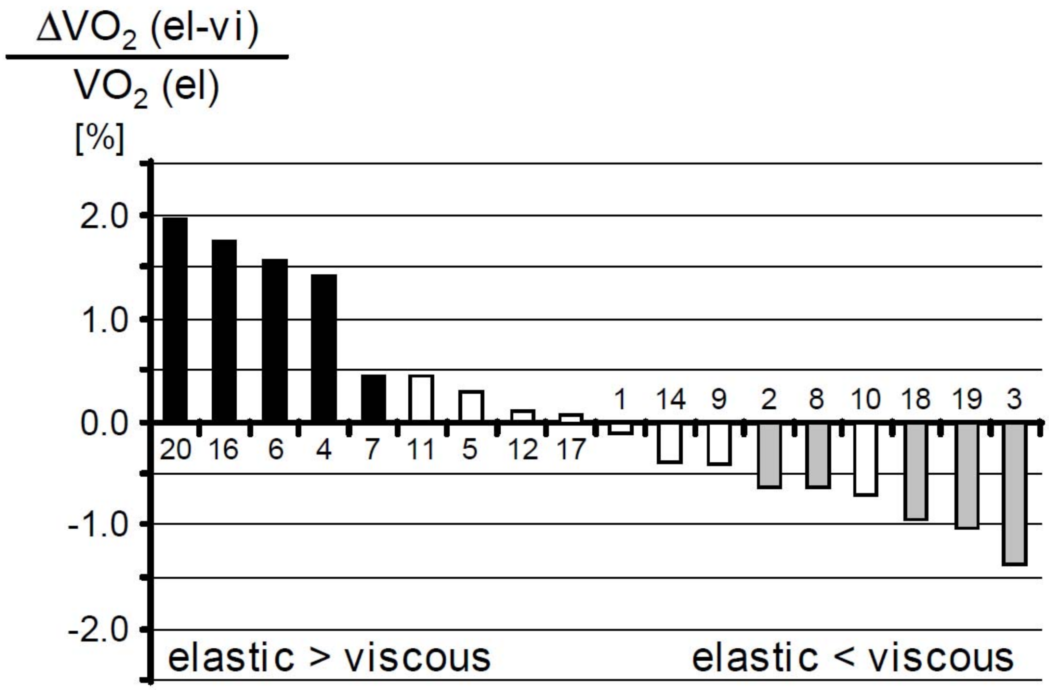


Figure 7: Subject-specific trends of metabolic expenditure while running on elastic and viscous midsoles. Each numbered column represents a different subject's comparative metabolic response (in percent change) while running on elastic versus viscous midsoles. Black shaded bars indicate significant ($P < 0.05$) trends towards increased metabolic expenditure on elastic shoes, grey shaded bars indicate significant trends the other way, and white bars indicate either no significant difference or no consistent difference across days. Only subjects that showed the same trends for all four trial sets (two reverse-order trial sets, two reverse-order experimental sessions) were shaded black or grey (Nigg et al. 2003).

A substantial amount of research on footwear cushioning has been performed for running (presumably fueled by the running shoe industry and competitive runners). However, less empirical data exists for walking, which was the focus of this thesis. The significant changes in running gait due to alterations in the properties of the contact surfaces give reasonable support to the idea that the body's kinematic and kinetics may change due to similar alterations during walking gait.

1.4.4 Surface and Shoe Stiffness (Walking)

In order to analyze walking gait, it is important to know how walking has been modeled in recent literature, how well these models are supported by experimental evidence, and what predictions these models make as to how the body's kinetics would change from changing shoe and surface parameters. As seen in Section 1.4.3, many experiments have been published concerning cushioning effects of running, but fewer have been published for walking, and those that have been published focus not on walking power and work but on other primary metrics like ground reaction forces and tibial accelerations, relating cushioning to lower-extremity shock and strain (Lafortune and Hennig 1992, Voloshin and Wosk 1981).

The single-support phase of the walking gait cycle has been compared to an inverted pendulum, and this model of walking captures many important features of human gait (Cavagna et al. 1976). Other models of walking have built upon this model by adding swing leg mechanics (Mochon and McMahon 1980). The authors who proposed that model stated that the "forces and the angle [of the leg in the inverted pendulum model] all execute trajectories which are quite close to those of the ballistic walking model…" (Mochon and McMahon). The inverted pendulum model will be the model referenced in the remainder of this thesis.

When walking like an inverted pendulum, the model predicts that for the duration of single support (i.e., the inverted pendulum arc), kinetic energy is converted into gravitational potential energy and then back to kinetic energy as the body's COM moves perpendicular to the support leg (Cavagna et at. 1977). During this phase of walking, theoretically, no mechanical work need be expended by the body through muscle work. The energetic cost of locomotion in this manner, however, is exacted as the body switches from one inverted pendulum arc to the next during the step-to-step transition at double support (see Figure 8). Here, positive work must be provided by the lower extremity musculature to offset the mechanical energy lost at collision (Cavagna et al. 1976, Cavagna et al. 1977, Donelan et al. 2002a, Donelan et al. 2002b, Kuo et al. 2005). In fact, it has been shown that the phase of walking associated with the largest generation of positive COM work is push-off, which occurs during double support (Kuo et al. 2005). Figure 9 shows this calculation across many walking speeds. Using the individual limb method (Donelan et al. 2002a), positive and negative power from each leg were integrated separately over a stride and just over collision and push-off, and the resulting work was divided by stride period. At slow to moderate speeds (<1.75 m/s in Figure 9), the positive power provided during double support (unfilled squares) makes up the majority the positive power over the stride (filled squares). A similar trend is found with the collision power over a stride (filled circles) and the total negative power over a stride (unfilled circles).

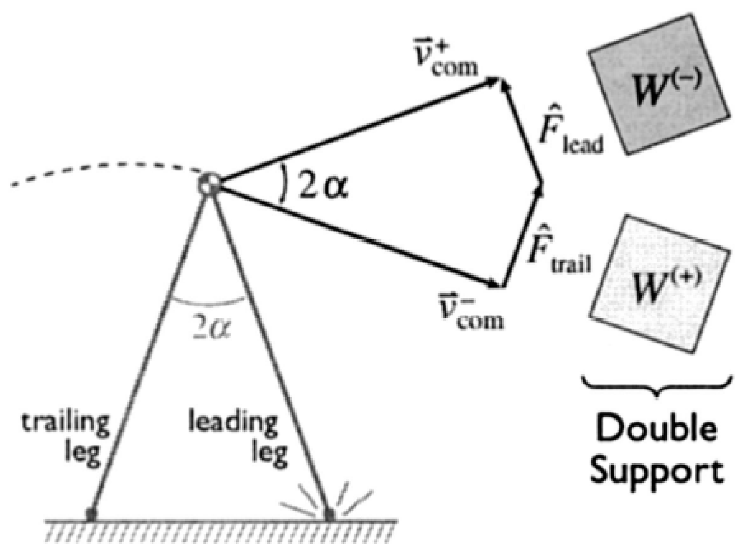


Figure 8: Simplest inverted pendulum model of walking, highlighting double support. In this model, push-off and collision occur simultaneously, minimizing required positive work to maintain steady walking gait (Kuo et al. 2005).

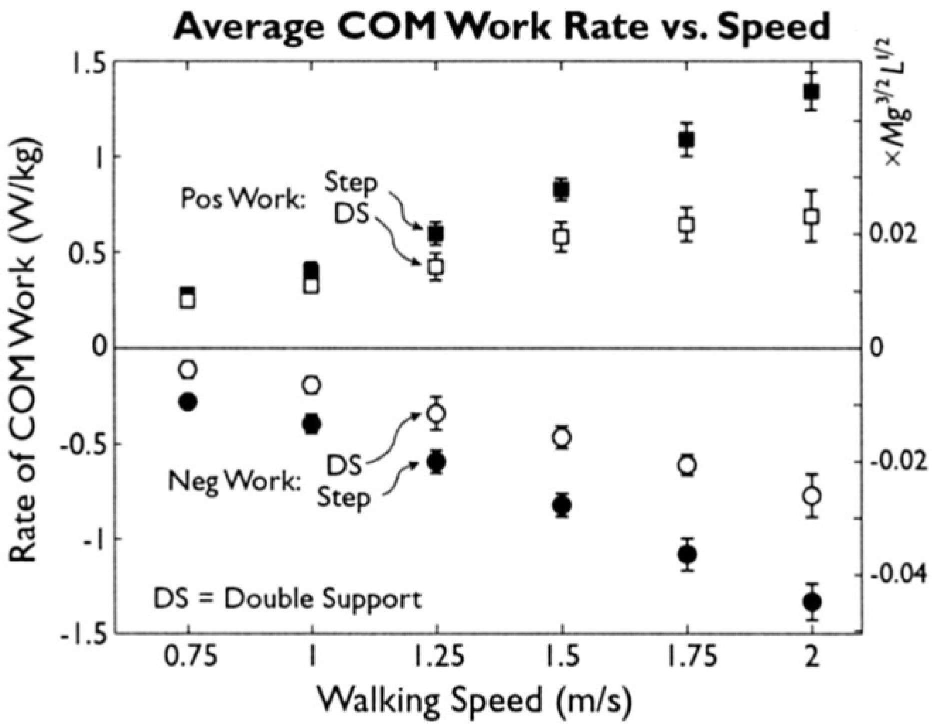


Figure 9: Average COM positive and negative power and double support positive and negative power at various speeds. As can be seen, positive COM power is primarily generated during double-support for slow and moderate (<1.75 m/s) walking speeds (Kuo et al. 2005).

During constant speed, level walking, total negative collision work done on the COM by the leading leg should theoretically cancel out (for model-predicted optimal gait) the total positive push-off work done by the trailing leg (Kuo et al. 2005). During phases of average walking speed increase or uphill walking, push-off work should be greater, and during phases of average walking speed decrease or downhill walking, collision work should be greater.

The model has also been used to characterize walking at different step lengths and step widths. Calculations based on the model predict that increasing step length exacts a COM work cost (Donelan et al. 2002b). Increasing step width is predicted to exact a similar cost (Donelan et al. 2001). Both these trends have been observed experimentally (Donelan et al. 2002b, Donelan et al. 2001) for COM work calculations and metabolic cost measurements.

Though the inverted pendulum model performs well for predictions of COM work and velocity during the different phases of walking, by design there are some measures it cannot predict and some factors it does not take into account. One of the metrics unpredicted by inverted-pendulum walking is the time duration of COM redirection. In the model, work performed by the trailing and leading legs during double support occur at small, but undetailed durations. This metric has been measured in other studies to date (Adamczyk and Kuo, 2009). Furthermore, the inverted pendulum model makes some assumptions, namely the rigid leg assumption, that are not true for the human body. One of the resulting consequences of this assumption is, again, that COM redirection does not occur instantaneously, and in fact takes a predictable amount of time (or percent stride) (Adamczyk and Kuo, 2009).

Mapping between the biomechanical realm and the metabolic realm is difficult at best and not well understood, and some aspects of metabolic concerns in walking are not well represented in any mechanical models. One example of this difficulty is that swing phase in human gait is not truly predicted to exact zero metabolic cost (Neptune et al. 2004, Umberger 2010). Since the knee generally is not at full extension during a given stride and the metabolic cost of supporting body weight is not zero, stance phase is predicted by some (Neptune et al. 2004, Umberger 2010) to not only have a metabolic cost, but have one that may be comparable if not greater than the cost of double support.

Though many inverted pendulum models have been adapted to model various parameters, the simplest inverted pendulum models do not take into account deformation of soft tissue in the body. This issue becomes the most substantial during the collision phase of walking gait where the heel pad and other dissipative areas (including the midsole during shod walking) play an important role in mediating collisional effects. Furthermore, soft tissue deformation sometimes has the potential to actually store energy and return it, a phenomenon sometimes referred to as elastic rebound (Kuo et al. 2005, Zelik and Kuo 2010), and this effect is left unaddressed by the simplest inverted pendulum models. Finally, the simplest inverted pendulum models are unable to quantify stored tendinous elastic potential energy, the release of which can make up a significant portion of push-off work (Sawicki et al. 2009, Zelik et al. 2014).

Additions to the inverted pendulum model of walking (Zelik et al. 2014, Dean and Kuo 2009) have been shown to alleviate a few of the limitations of the simplest model alone. The addition of a passive spring akin to the Achilles tendon and a foot segment of specified length can be show to, in the ideal scenario, reduce the needed work of active components (plantarflexors of the ankle) to zero (Zelik et al. 2014). Though this model still has a large body of assumptions that the human body does not match (for example, the Achilles tendon cannot act in a passive manner without plantarflexor muscle activation in some regard, even when just providing a clutch-like mechanism), it still shows that walking can, with appropriate modeling, be a lossless task.

1.5 Measuring Push-Off and Collision Phases of Walking

The phases of walking where changes in COM, joint, and deformable foot powers, resulting from changing sole cushioning, will likely show the most substantial variability are push-off and collision. COM power is typically plotted against percent stride (see Figure 10 for an example), and this curve, the push-off phase is defined as the region between the crossing of zero power around 50% stride and the return to zero around 60-70% stride (where the toe leaves the ground, or toe-off). The collision is defined similarly as the region between 0% stride (where the heel makes first contact with the ground, or heel-strike) and where the curve crosses zero near 15% stride.

Impulsivity of push-off and collision events affect the energetics of walking. Many studies (Ruina et al. 2005, Kuo et al. 2005, Yeom and Park 2011) discuss the impulsive nature of push-off and collision. The time it actually takes to go through the step-to-step transition has an important effect on the model predictions for various metrics. One key finding is that the more time it takes, the more gravity has an effect on the momentum change of the COM (Yeom and Park 2011). That means, based on this gravity-inclusive walking model, that a shorter, more impulsive step-to-step transition is energetically more ideal. An easy way to compare COM power data between trials then is to estimate impulsivity by comparing peak values in the push-off phase and peak negative values in the collision phase. This technique will be used as a main method of comparison between trials for the experiment presented in this thesis.

1.6 Hypotheses

The experiment presented in this thesis was conducted with three hypotheses in mind. The first hypothesis was that the effects of heel cushioning and forefoot cushioning on collision and push-off respectively could be decoupled. In other words, if, for example, soft heel cushioning was determined to benefit subjects in some quantifiable way but hard forefoot cushioning was also determined to be beneficial in another way, a shoe with both these qualities would maintain the beneficial nature on collision and push-off without losing much through interdependence issues. This was postulated because the stance phase of walking gait in healthy individuals typically begins with foot strike, which occurs exclusively at the heel, and ends with push-off, which occurs exclusively around the ball of the foot and toes. Since, theoretically, little work needs to be done during weight transfer to the front of the foot in the rebound and pre-load phase, it is logical to predict that transitioning between different sole properties at the heel and forefoot during those phases would not have significant energetic effects.

The second hypothesis was that a softer forefoot cushioning condition would result in smaller push-off peak power. This prediction was based in the fact that softer forefoot cushioning was expected to absorb more power, similar to how sand absorbs more power when walking along the beach than does a hard walking surface, taking away from the ability of the biological system to provide positive power to the ground.

The third hypothesis was that soft heel cushioning would result in a larger magnitude peak power during collision. Since softer heel cushioning was predicted to absorb power on top of the typical power absorption by the body, the effects were expected to sum. The corollary to this hypothesis is that the absorptive power provided by the body proximal to the foot would be smaller. This prediction was rooted in the cost of cushioning hypothesis (Frederick et al. 1983) extrapolated to walking, as well as further evidence that surface cushioning can absorb collision energy, leaving less for the body itself to absorb (Skinner et al. 2015).

CHAPTER 2

METHODS

2.1 The Development of a Testing Platform

The first and arguably most crucial component of this experiment was the design of a test shoe that would allow for the swapping of midsoles quickly to facilitate a large number of testable conditions for a given experimental session. The design requirements to be met can be seen in Table 1.

Table 1: Design Requirements for Testing Shoe

Mass	Whatever means by which variable midsoles will be attached must have a negligible mass compared to the mass of the shoe and the mass of the attached midsole
Speed	Whatever means by which variable midsoles will be attached must allow for a "quick" swapping of conditions
Base	Whatever shoe is chosen to be the base to which conditions will be attached must have near-negligible midsole cushioning compared to the conditions or must be altered to satisfy that requirement

The solution produced during brainstorming to best satisfy all design requirements was to find a base shoe with thin to negligible sole cushioning properties and attempt to use standard hook-and-loop fastener with adhesive to temporarily attach the experimental conditions. A base "skate sneaker" shoe (Etnies, Lake Forest, CA) was selected that had only very mild midsole cushioning compared to the experimental conditions (see Section 2.2) and was relatively flat. The flatness was deemed desirable in order to better facilitate the firm attachment of the cushioning (see Figure 10). The insole was removed from the shoe to further decrease the nominal cushioning.

Figure 10: Base shoe for testing platform (Figure retrieved from Amazon.com).

The next step of fabrication was to adhere hook (i.e., hook-and-loop) fabric to the existing outsole, adhere loop fabric to a sample of soling material cut to match the shape of the shoe's outsole, and test the strength of hold during extended walking and jogging tests interspersed with periods of running. It was found that a convenient adhesive which allowed for an adequately durable hold was high performance acrylic hot glue (Surebonder-FPC Corporation, Wauconda, IL). After the hook-and-loop fastener was properly applied to the shoe and test sole, it was tested as described above. Qualitatively, the performance of the hook-and-loop was more than adequate for this application. It held the applied sole both vertically (i.e., the sole did not separate from the test shoe, even as the shoe bent near the metatarsophalangeal joint) and anteriorly/medially (i.e., the sole did not translate along the test shoe in any appreciable way).

Even during periods of running, the strength of hold did not seem to degrade. The test was a success, and the hook-and-loop method of attachment satisfied all three listed design requirements (see Table 1) to a greater extent than any other produced idea feasibly could.

2.2 Experimental Condition Design

The next important component of this experiment was to fabricate the experimental conditions. Soling material (SoleTech, Salem, MA) was acquired through a distributor (Cascade, Chico, CA) with manufacturer-provided Shore A durometer ranges of 17-20 (Ultra Cloud) for the softest condition, 35-40 (Cloud) for the medium condition, and 60-65 (SBS Firm) for the hardest condition (soletech.com). These conditions will be referred to as the soft, medium, and hard conditions for the remainder of this thesis. The soft and medium conditions are made of Ethyl Vinyl Acetate (EVA), a common midsole material, while the hard condition is made from a blend of EVA and Styrene Butadiene Rubber (SBR, also known as synthetic rubber). These materials were chosen to sweep a wide range of possible durometers (~20-60 Shore A) in a somewhat linear fashion. The thickness of each cushioning sheet was chosen to be 12.7 mm (0.5 inches) in an attempt to ensure the trends between the cushioning properties is appreciable.

From these sheets of soling material, silhouettes were cut out to match the profile of the base shoe's outsole. From there, the silhouettes were cut roughly just beyond the most anterior point in the heel pad (see Figure 11). This was done to test if effects of the sole cushioning at the heel can be isolated from the effects of the sole cushioning at the forefoot, and if so, what the effects of cross-conditions are.

Figure 11: Sole cushioning conditions showing color, thickness, heel-forefoot separation, bending slits, and adhered loop (for hook-and-loop attachment). From top to bottom: Hard forefoot and heel, medium forefoot and heel, soft forefoot and heel.

In order to ensure the cushioning was substantial enough to tease out possible trends from the experiment, the cutouts were made in sets of three such that two could be adhered together to form a doubled-thickness condition (25.4 mm thickness) for each level of durometer, soft, medium, and hard. These different thickness conditions will be referred to as short (single thickness) and tall (double thickness) when not specifically referenced by thickness measurement.

In order to isolate sole hardness from bending stiffness, horizontal slits were cut into the foam soles at regular intervals (roughly 1.5 slits per 10 mm for a range of 38 mm (1.5 inches) in front of and 38 mm (1.5 inches) behind the metatarsophalangeal joint) to a depth such that the bending stiffness in the direction of dorsiflexion was qualitatively close to that of the softest condition (see Figure 11). In order to test the validity of this alteration, one condition (hard forefoot, hard heel, short) was repeated without slits and included in the experiment (referred to as the stiff condition).

2.3 Experimental Protocol

Eight (N=8) subjects were tested. All were male, mean age 22.2 years (1.39 years standard deviation), mean height 178.4 cm (5.72 cm standard deviation), mean mass 77.9 kg (8.50 kg standard deviation) with mean size 10.5 (mean 10.6, 0.75 standard deviation) shoes. The exclusion criteria were not being able-bodied, not having continued complications from any previous surgeries on the lower limbs, and not having feet correctly sized to fit into the testing shoe. Since the test shoe was men's size 10.5, individuals with men's size 9.5-11.5 shoes were considered a fit.

Before subject arrival, consent forms (this study was approved by Vanderbilt University Institutional Review Board) were prepared, data collection software and hardware was prepared, and motion capture hardware was calibrated. The treadmill used in this experiment was Bertec's Fully Instrumented Treadmill (Bertec Corporation, Columbus, OH), a split-belt treadmill equipped with two sets of piezoelectric force plates capable of measuring forces, moments, and centers-of-pressure (COPs) under each foot during locomotion. The motion capture system utilized in this experiment was a Vicon ten camera system (Vicon Motion Systems, UK), and data collection occurred through Vicon's capturing software, Nexus. Calibration was performed no greater than an hour before subject arrival.

Upon arrival, subjects were briefed as to the pertinent information about the experiment and were asked to sign a subject consent form after identifying and discussing the potential risks associated with participation in the study. In one case a media release form was signed as well. A sample consent form can be seen in the Appendix. They were then asked to change into Spandex shorts in order to facilitate good tracking for motion capture markers that would have to be placed on the upper thigh (see Appendix for marker placement image). Then, a series of markers were placed at various bony prominences on the lower extremities and pelvis for later use in kinematic and inverse dynamic analysis. Marker clusters were also placed on the segments (i.e., the thigh and shank) in groups of four, ensuring they were non-collinear and spread out along the segment and placed to avoid skin movement artifacts as much as possible (Cappozzo et al. 1997). For a complete list of markers and a figure showing them placed on a subject, see the Appendix.

The subjects were then asked to step onto the treadmill for a static calibration, followed by a dynamic calibration, both to be used during marker processing and again during data analysis through a standard 3D analysis toolkit, Visual3D (C-Motion, Germantown, MD). Safety harnesses mounted to structural beams were worn by subjects at all times during testing and calibration in case of a fall. The treadmill was then brought to a speed of 1.0 m/s for a one minute warm up, followed by another one-minute warm up at 1.4 m/s, the speed at which the experimental trials would be measured. Since the subjects were barefoot at this point after the calibrations, they were asked to come off the treadmill and put on the testing shoes. The reason a warm up occurred barefoot is because this experiment was conducted at the same time as a related experiment that contained barefoot trials. At this point the testing shoes had a control condition of Vibram (Albizzate, Italy) outsole attached. Warm up with the shoes for one minute at 1.0 m/s and another minute at 1.4 m/s was then completed. For the last of these warm up trials, ground reaction force (GRF) data was collected and fed through a custom-made MATLAB (MathWorks, Natick, MA) script to find average step frequency.

At this point the experiment began. The conditions were pseudorandomized for each subject such that the order of the cushioning conditions on the forefoot was randomly selected, and the order of the heel cushioning for any given forefoot condition was randomly selected. This way, the forefoot cushioning could be left on for two thirds of condition swapping instances, reducing total experimental session duration. Breaks from the experiment were offered based on subject need. Table 2 shows the set of total experimental conditions for the short trials before pseudo-randomization. The same set of conditions was present for the tall trials, excluding the stiff condition.

Table 2: Set of All Experimental Conditions for Short Trials

Forefoot	**Heel**	**Forefoot**	**Heel**
Hard	Hard	Medium	Soft
Hard	Medium	Soft	Hard
Hard	Soft	Soft	Medium
Medium	Hard	Soft	Soft
Medium	Medium	Stiff	Hard

GRF data was collected at 2000 Hz and motion capture data was collected at 100 Hz for three consecutive 30 second intervals from the moment the treadmill accelerated up to 1.4 m/s. The data from the third trial for each condition set, barring issues that may have arisen, was the data selected to process further; the preceding trials were captured as well for redundancy purposes. By the time data from the third trial of the first condition set was taken, the subject had warmed up on the treadmill for a minimum of five minutes (corresponding to roughly 400 steps), which has been shown to be the minimum necessary warm up time for treadmill locomotion familiarization (Owings and Grabiner 2003, Zeni Jr. and Higginson 2010). During a subset of experimental conditions (all short combinations with medium heel or forefoot, as well as the hard-hard and soft-soft conditions), two extra 30 second data collection periods occurred where the subjects were told to match pace with a metronome set to their self-selected step frequency measured from warm up trials.

In order to avoid changes in gait kinetics and kinematics due to velocity, the entire experiment was conducted at 1.4 m/s. In order to avoid changes to changing mass characteristics between trials, masses were added to the top of the testing shoe via adhesive-backed hook-and-loop fastener. Table 3 shows the weight added to the test shoe given each heel and forefoot condition. As can be seen, the conditions were weight-matched to the heaviest conditions (tall, hard condition for both the heel and forefoot). Since both velocity and shoe mass were matched across all trials, kinematic and kinetic metrics can be directly compared.

Table 3: Masses Added to Each Condition for Weight-Matching

	Heel	Added Mass (g)	Forefoot	Added Mass (g)
Short (12.7 mm)	Hard	28	Hard	75
	Medium	35	Medium	102
	Soft	41	Soft	122
			Stiff	75
Tall (25.4 mm)	Hard	0	Hard	0
	Medium	22	Medium	62
	Soft	28	Soft	91

2.4 Data Analysis

Marker processing and gap filling was performed in Vicon Nexus 2. From there, C3D files were uploaded into Visual3D for data compounding and export. In order to export joint-based kinematic and kinetic data, functional joints (Schwartz and Rozumalski 2005) were found through the dynamic calibration motion files. The filtering of the marker data utilized a Butterworth filter with a cutoff frequency of 6 Hz (Winter et al. 1974). The GRF data was filtered the same way with a cutoff frequency of 25 Hz. Outputs of this process were joint angles, moments, and powers (3DOF rotational power); filtered GRFs, COPs, and free-moments; and foot segment COM locations, velocities, and angular velocities. This data was all fed into a custom-written Matlab script to perform COM power and work calculations (Donelan et al. 2002a) and deformable foot power calculations (Siegel et al 1996, Takahashi et al. 2012, Takahashi and Stanhope 2013), and to parse out the data into strides and average all metrics across strides. Inter-subject averages for applicable metrics were also calculated.

2.5 Statistical Analysis

For comparisons between either heel or forefoot cushioning conditions (i.e., comparing hard-hard to hard-med to hard-soft) for all metrics, an analysis of variance (ANOVA) algorithm was implemented with significance defined at $P < 0.05$. Where the ANOVA test showed statistical significance, post-hoc pairwise comparisons (three pairwise comparisons per test) were made using the Holm-Sidak test (Cardillo 2008). For comparing between the hard and stiff forefoot cushioning conditions for all metrics, a paired t-test was employed with statistical significance defined at $P < 0.05$. A statistical power calculation was performed as well.

CHAPTER 3

RESULTS

3.1 Effects of Varying Forefoot Cushioning on Power

Before presenting those findings however, it is important to present a typical COM individual-leg power curve highlighting exactly where push-off is represented. The plotted curve in Figure 12 shows a standard gait cycle represented through individual-leg COM power. In this case, the shaded area under the curve highlights the push-off phase of walking gait, where the trailing leg adds energy to the COM just before and during collision. A stride is defined as one heel-strike (0% stride) to the next same-leg heel-strike.

It was found that forefoot cushioning indeed had a significant effect on various metrics measured in this experiment. In Figure 13, the COM power curves of a representative subject for three different trials are plotted. The short thickness between these trials was unchanged, as was the soft heel condition. The only variable altered between the three trials was the forefoot cushioning condition. The soft forefoot condition resulted in the highest peak power magnitude, followed by the medium forefoot and finally the hard forefoot. The peak magnitudes, normalized to subject mass, were 3.27 W/kg, 3.08 W/kg, and 2.93 W/kg, respectively. The soft forefoot condition then resulted, for this subject, in an 11.6% increase in COM push-off peak power for an average stride. This result alone signifies that forefoot cushioning has a substantial effect on the kinetics of walking for at least some people, but it is very important to determine if the effect is the same for all subjects, and whether or not the effect is statistically significant.

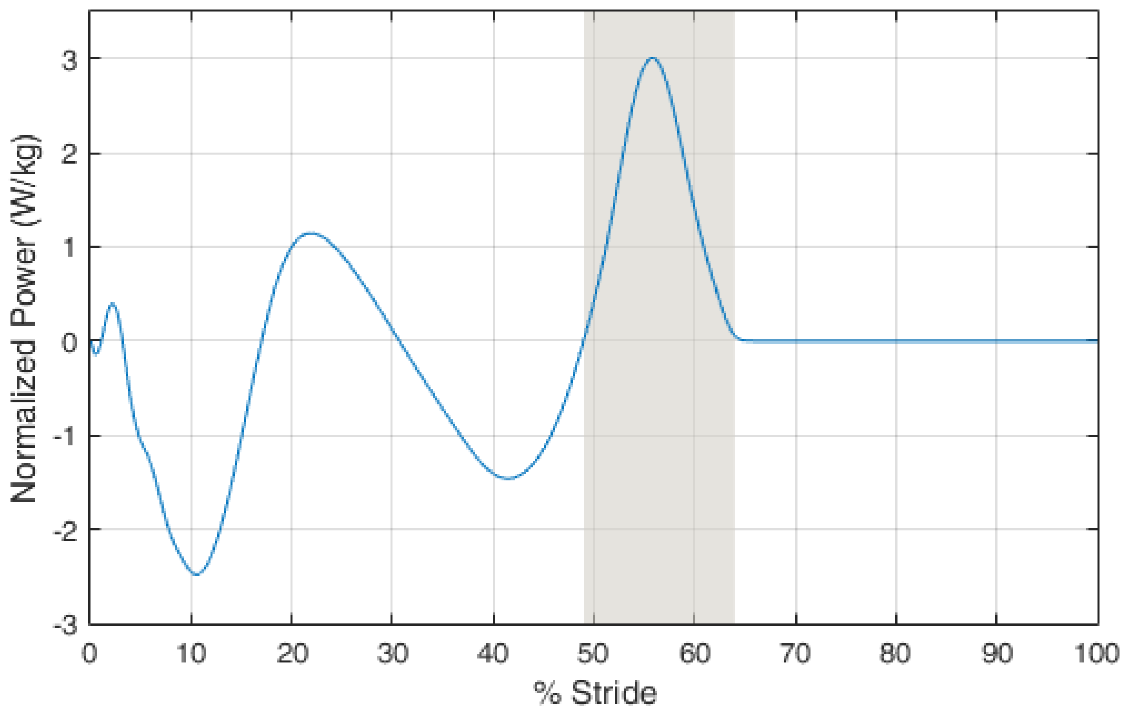


Figure 12: Typical COM power plot for a single leg with push-off phase highlighted. Power in this case is normalized to subject mass, and the curve is plotted against % stride. In all plots, a stride is defined from foot contact (heel strike) to subsequent ipsilateral foot contact.

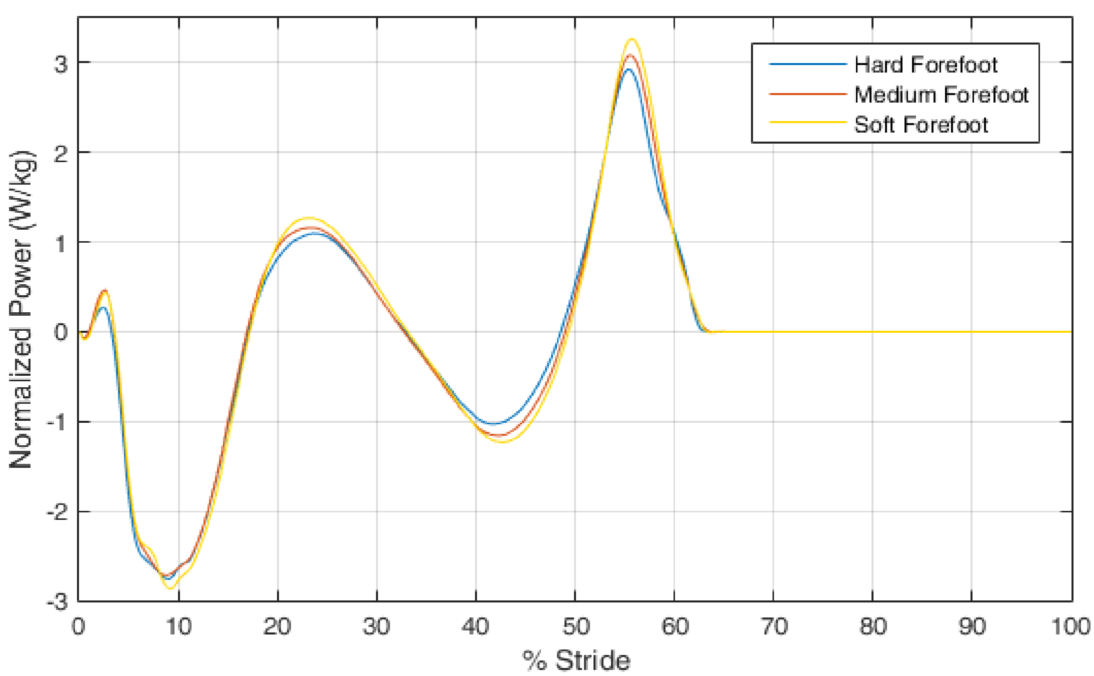


Figure 13: COM power plots for a representative subject varying only forefoot cushioning. These trials were all short conditions, and the heel cushioning was soft. The plots are the average COM power (computed from 30 seconds of strides for both legs).

Indeed, a significant increasing trend was found for the COM peak power magnitude and forefoot cushioning condition across subjects, thicknesses, and heel cushioning conditions. Figure 14 shows a representative plot of this trend, averaged across subjects, with bars representing the standard deviation. For this thickness and particular heel condition (short, soft heel), the values spanned a range from 2.85 W/kg to 3.10 W/kg, a difference of 0.25 W/kg. The percentage increase of the COM push-off peak power (comparing the soft forefoot to the hard forefoot) was 8.8%. Statistical analysis on this set of conditions yielded significance between the hard and soft forefoot conditions (P = 0.0003) and between the hard and medium forefoot conditions (P = 0.017).

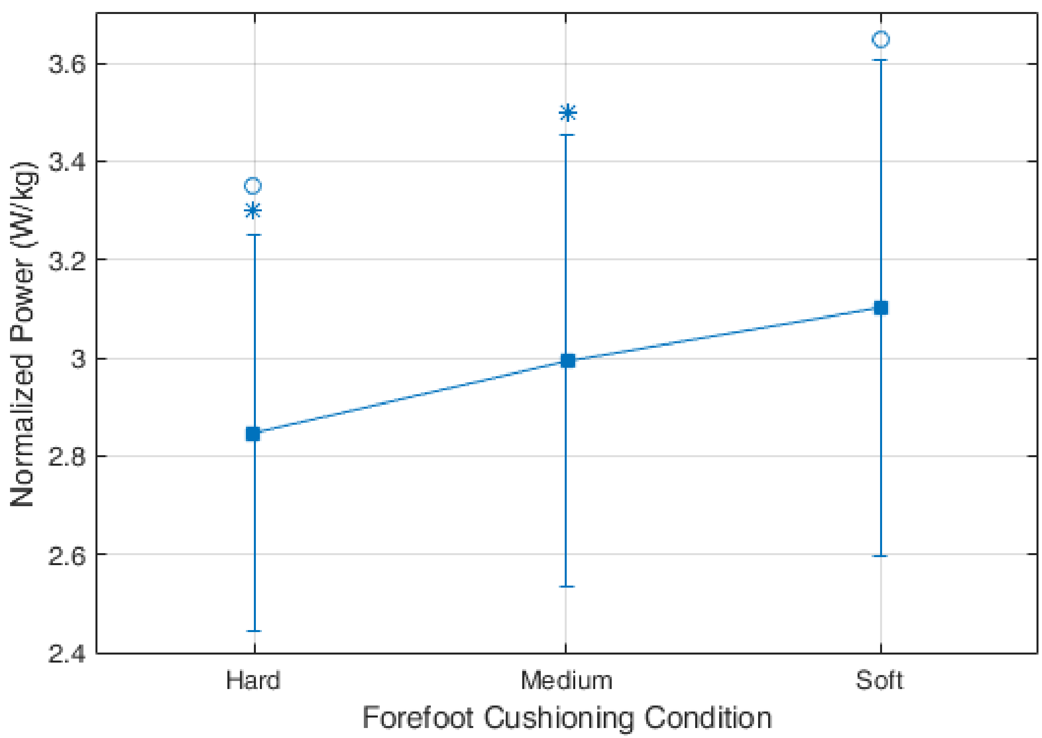


Figure 14: COM push-off peak power values varying forefoot cushioning. In this representative plot, the thickness was held fixed at 25.4 mm and the heel was soft. The '*' and 'o' represent significant pairs.

This trend of increasing push-off power with softer forefoot cushioning, when averaged across subjects, was found for both thicknesses and all heel conditions. Figure 15 highlights this finding by plotting each curve (without standard deviation bars for clarity) as an increase above the peak power value for the hard condition. In this way, vertical shifts of the data can be neglected and the trends between hard, medium, and soft forefoot cushioning can be emphasized. The statistical significance presented in the Figure was analyzed before shifting all curves. For further reference, the light blue curve in this Figure is the same data displayed in Figure 14.

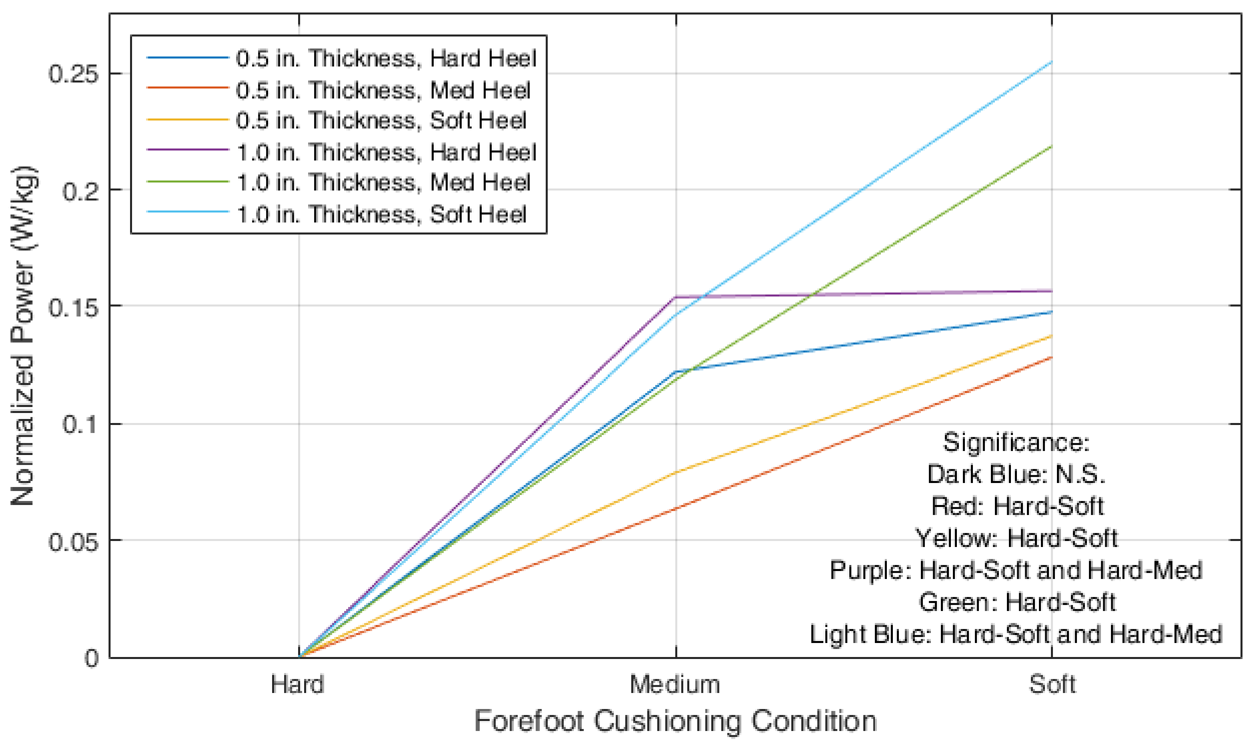


Figure 15: COM push-off peak power for all combinations of heel cushioning and thicknesses as differences from hard forefoot condition. The trends found when moving towards softer forefoot cushioning are similarly increasing for all curves shown. Statistically significant pairs are shown in the text at the bottom-right of the figure. The only plot without significance is the short, hard heel comparison. Standard deviation bars were omitted for clarity.

A key finding is also that changing forefoot cushioning did not significantly affect COM collision power. Figure 16 is plotted with the same scale on the power axis as was Figure 14 for easy comparison, and illustrates this result.

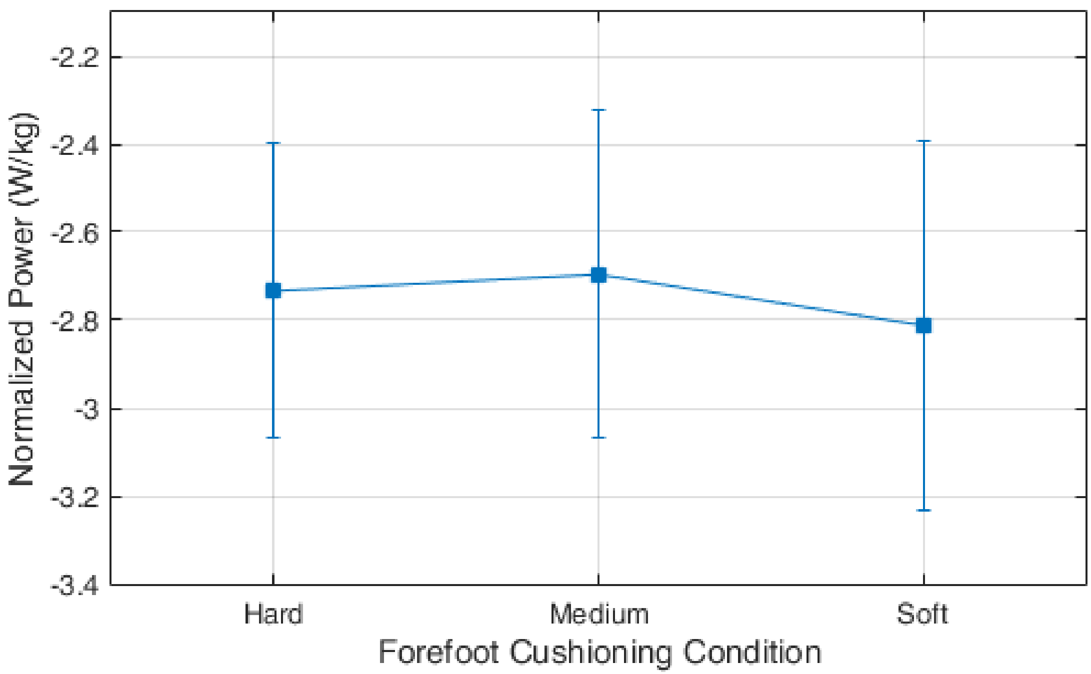


Figure 16: COM collision peak power magnitude varying forefoot cushioning. No significant difference (P = 0.37) was found between these values, and the magnitude difference across the largest range of values is only 0.11 W/kg, representing only 4.1%. This representative plot was for the same short, soft heel plotted in Figure 14.

In order to tease out the specific causes of the trend found in COM push-off peak power, the location along the percent stride axis was determined for the peak COM push-off power for each condition for each subject, and the three degree-of-freedom (3DOF) power curves for the ankle, knee and hip were probed at that location. The foot deformable power was also probed at this location. This way, the components of joint and foot power that may make up the observed trend in COM power can be analyzed, even if peaks in joint-level and foot power do not well align with COM power.

First, however, it is important to note the similarities between 3DOF power plus foot power (referred to for the rest of this thesis as 3DOF+F power) and COM power. As can be seen in Figure 17, the trends and magnitudes of these curves line up relatively well, but as discussed in Zelik et al. (2015), the curves are not expected to align without augmenting the COM curve to include peripheral powers (lower extremity motion relative to the COM) and performing a more rigorous six degree-of-freedom (6DOF) analysis, and even then collision is not fully accounted for (Zelik et al. 2015).

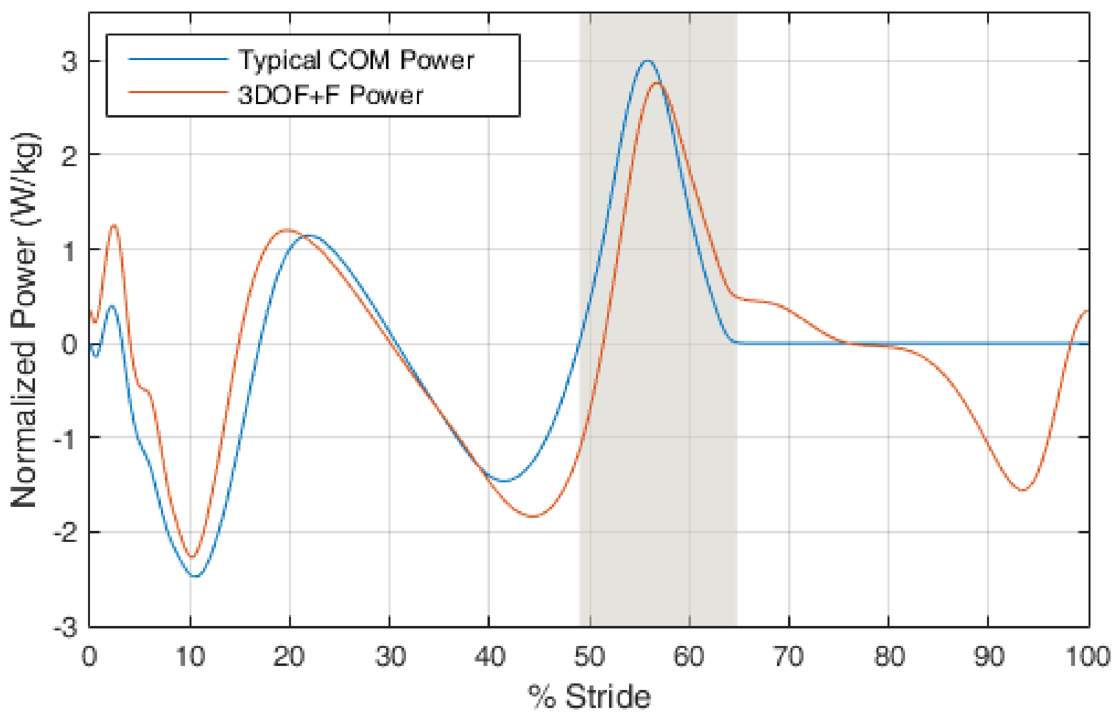


Figure 17: Typical COM power and 3DOF+F power plots with the push-off phase highlighted. The non-zero behavior exhibited by the 3DOF+F power curve is due to the power necessary to move the lower extremity segments with respect to the COM. Since this occurs during swing phase for the leg in question, it is not captured by traditional COM power curves.

Importantly, the peak power from 3DOF+F analysis is a known determinant of COM power since biomechanically, all positive power originates from muscle action or elastic return of stored energy. It is expected, then, that qualitatively similar trends should exist between COM power and 3DOF+F power at push-off. Indeed, as can be seen in Figure 18, this expectation is met. A quick glance at Figure 13 demonstrates that indeed the trends are similar between COM power and 3DOF+F power.

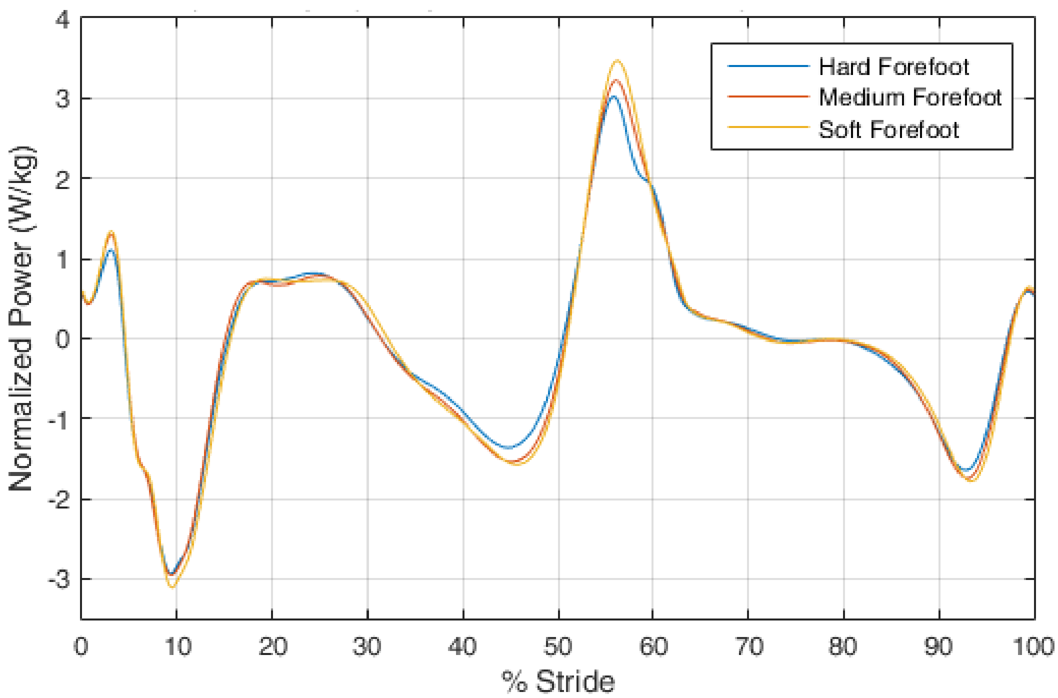


Figure 18: 3DOF+F power plots for a representative subject varying only forefoot cushioning. For these plots, the thickness was 12.7 mm, and the heel cushioning was kept as soft. The curves are plotted with the same scaling as is Figure 13 for easy comparison.

Next, the power magnitudes at COM peak push-off power for the hip, knee, ankle, and foot were examined in order to determine if, across subjects, there are any significant trends that might contribute to the trends found in Figures 14 and 15.

Figure 19 shows example trends in 3DOF+F power contributions from the hip, knee, ankle, and foot at the COM peak push-off power location. As can be seen, the only statistically significant trend was found for the ankle, which increases with similar magnitude from the hard to soft forefoot condition (0.16 W/kg) as the increases in COM power (see Figure 15).

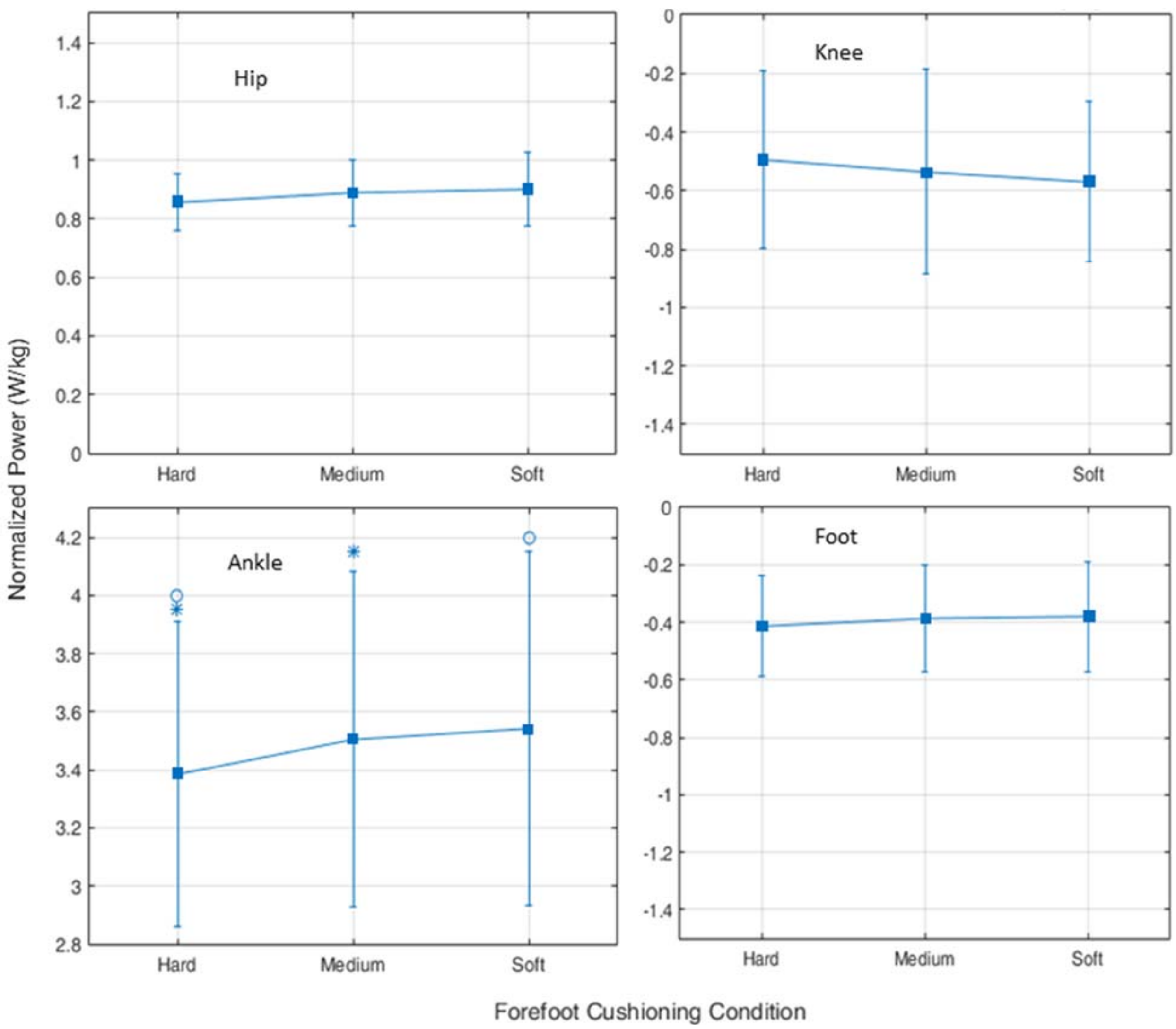


Figure 19: Contributions to 3DOF+F power at COM peak push-off power. The plots shown are representative of the trends seen across heel conditions for the short thickness. Clockwise from the top-left, the hip (short, soft heel) showed an insignificant trend, the knee (short, medium heel) showed an insignificant but decreasing trend, the foot (short, hard heel) showed an insignificant trend, the ankle (short, hard heel) showed a significant increasing trend. Significantly different pairs are shown with symbols '*' and 'o'.

The thickness seemed to play a role in how each joint was contributing power. Figure 20 shows a representative ankle plot for the tall trials. Interestingly, where the ankle shows the trend seen in Figure 19 for the short trials, in the tall trials it begins to show a less regular trend, increasing significantly (P = 0.009) from the hard to medium condition, and decreasing significantly (P = 0.002) from the medium to the soft condition.

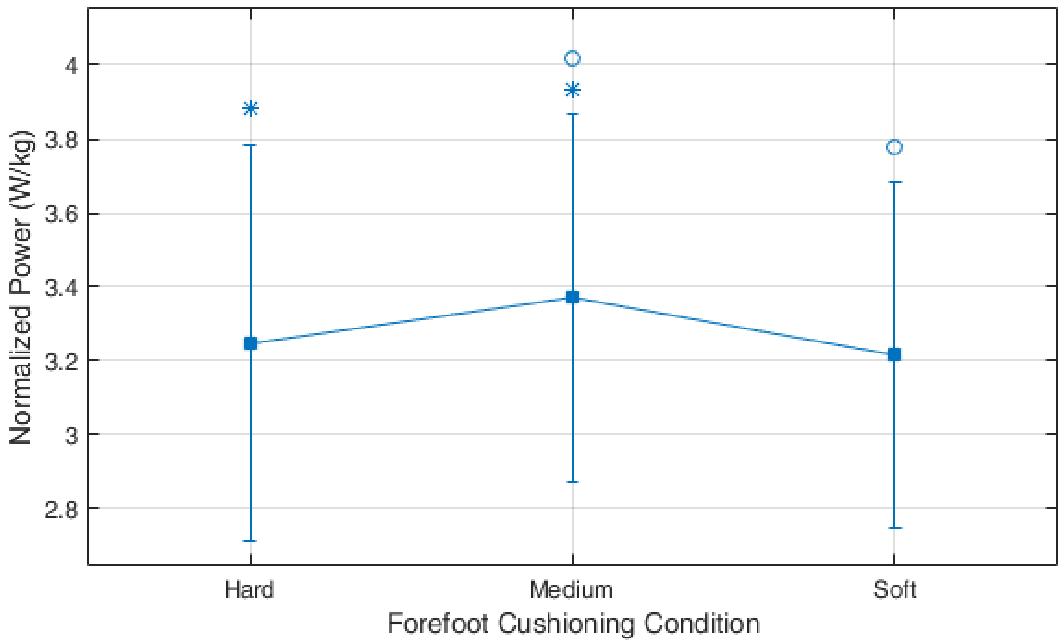


Figure 20: Representative ankle contribution to 3DOF+F Power at COM peak push-off power for thick trials. Significant (P < 0.05) pairs are designated by the symbols '*' and 'o'.

The subject-specific results are also worth noting. Figure 21 demonstrates the level of inter-subject variability found for tall, hard heel trial by plotting the data as the difference from the hard condition. It is interesting that though the trend is a significant (see Figure 15) increasing behavior from hard to medium and from hard to soft, some specific subjects did not follow this trend. Some showed a large increase from hard to medium and then little change from medium to hard as predicted by the overall trend. Some showed an initial increase from hard to medium, but a decrease from medium to hard. Some subjects showed a near linear increase from hard to medium to soft forefoot cushioning conditions, and one subject showed an initial decrease from hard to medium but little change from hard to soft.

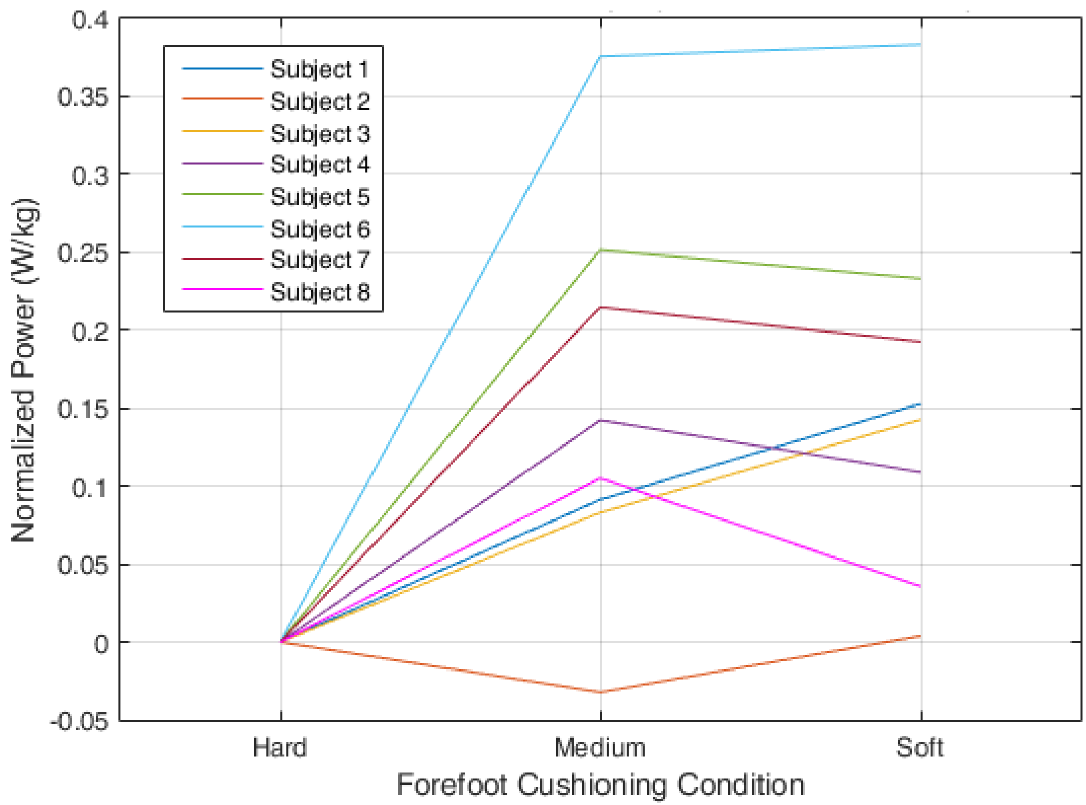


Figure 21: Hard-medium-soft forefoot peak COM push-off power comparison for all subjects. Note the general trend (refer to Figure 14 and 15), and the inter-subject variability in specific aspects of the trend. This comparison was made for the tall, hard heel trials across subjects.

3.2 Effects of Varying Heel Cushioning on Power

Similar comparisons can be made between heel cushioning conditions while forefoot and thickness effects are held constant. Figure 22 again shows a typical COM power plot for one leg, this time with the collision phase highlighted. In this phase, the leading leg touches down on the ground (known as heel-strike) and absorbs power, helping to redirect the COM to the pendular arc of the next step.

Changing heel cushioning while keeping other variables constant was found to have an effect on COM power curves, specifically in the collision phase, as would be expected. Figure 23 presents the COM power curves of a representative subject for hard, medium, and soft heel conditions, while the forefoot condition (soft) and the thickness (short) remained constant. The hard heel condition resulted in a peak collision power of -2.98 W/kg, the medium heel condition resulted in a peak collision power of -2.73 W/kg, and the soft heel condition resulted in a peak collision power of -3.22 W/kg. The maximum change, from the medium to soft heel, was -0.49 W/kg, which corresponds to a soft heel peak power that is 17.9% more negative than the medium condition peak power. It is noteworthy that an individual subject shows this trend, but it is important to check trends across subject averages.

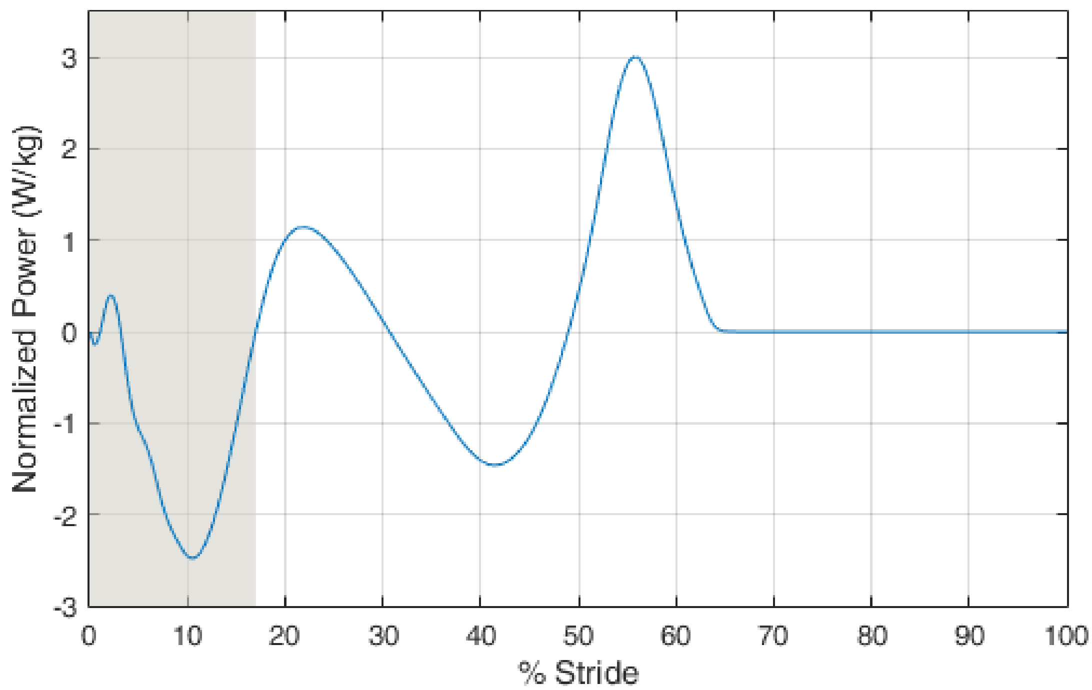


Figure 22: Typical COM power plot for an individual leg with collision phase highlighted. This data is plotted over a single stride.

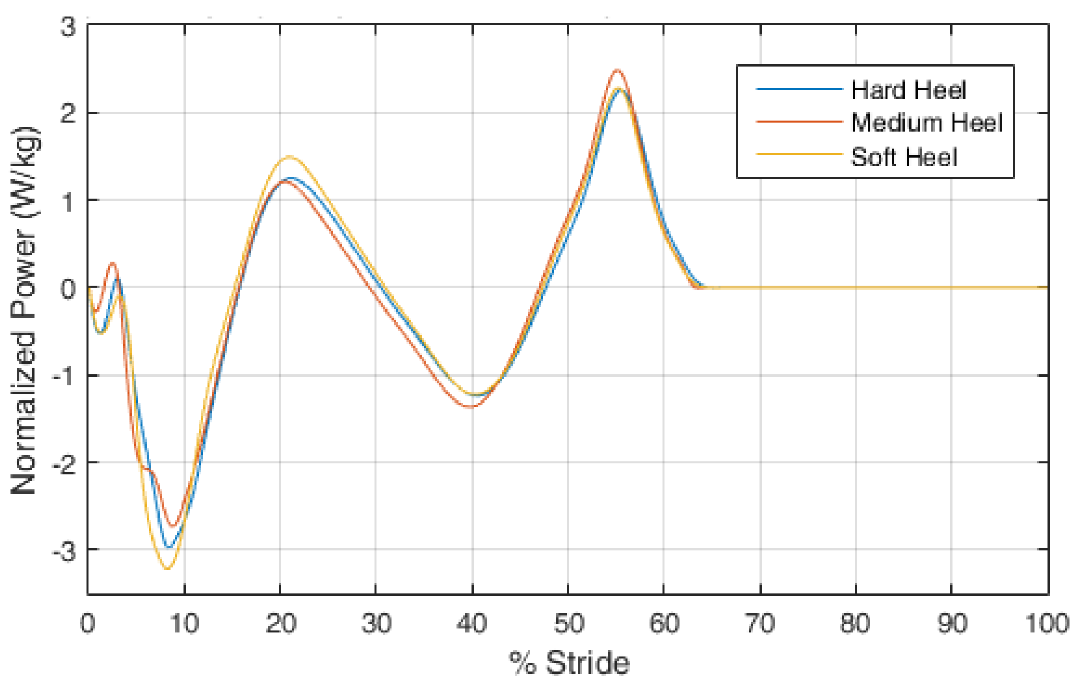


Figure 23: COM power plots for a representative subject for hard, medium, and soft heel conditions. The forefoot remained soft across each trial, and the thickness remained 12.7 mm for each trial as well.

The comparisons shown in the representative subject were also found across subject averages of COM peak collision power. Figure 24 presents this trend across subjects for only the tall, soft forefoot comparison for clarity of representation. The squares represent the means, the bars represent the standard deviations, and the other symbols signify statistically significant pairs. The COM peak collision power average was -2.54 W/kg for the hard heel, -2.50 W/kg for the medium heel, and -2.81 W/kg for the hard heel. After ANOVA, pairwise statistical significance was determined between the medium and soft heel conditions (P = 0.0003) and between the hard and soft conditions (P = 0.0009). The maximum magnitude increase (from the medium condition to the soft condition) was -0.31 W/kg, corresponding to a value 12.4% more negative for the soft condition.

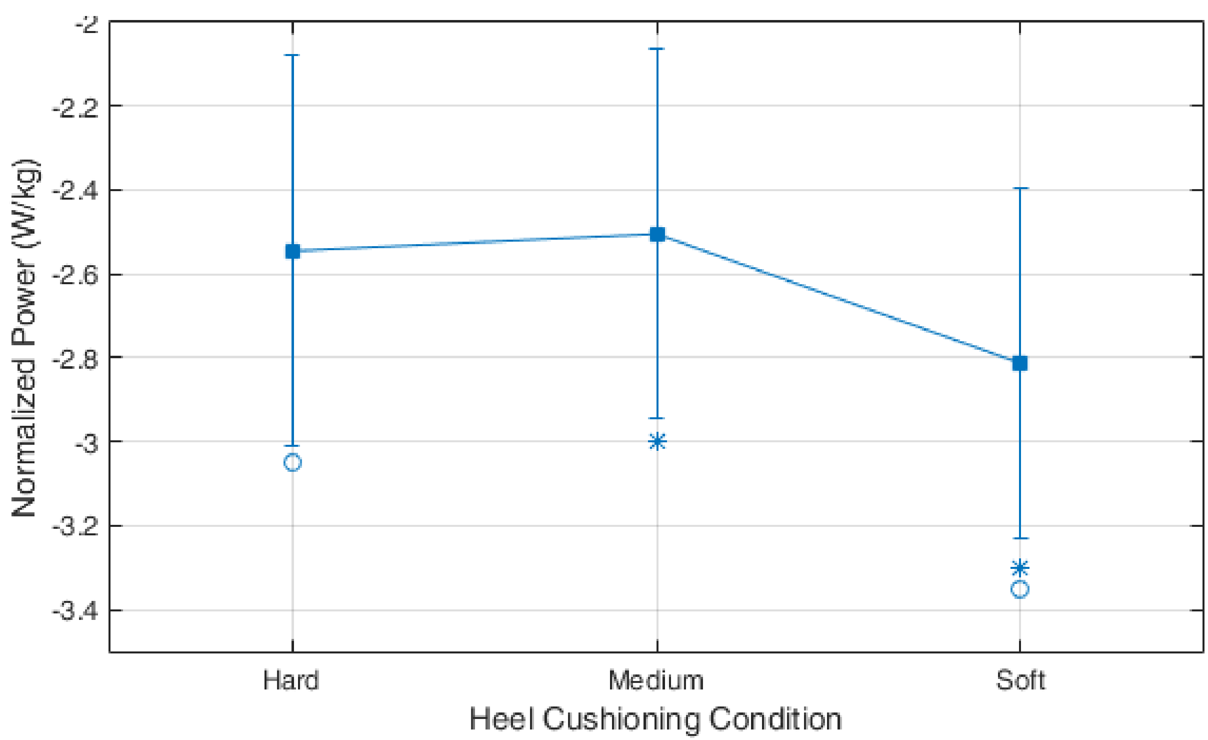


Figure 24: COM peak collision power averaged across subjects for varying heel cushioning. Thickness remained constant at 25.4 mm, and the forefoot cushioning was soft. Statistically significant pair groups are denoted by '*' and 'o'.

Figure 25 summarizes the effect of changing heel cushioning over all thicknesses and forefoot cushioning conditions. All hard-medium-soft heel cushioning comparisons yielded statistical significance: the medium-soft pair comparison was significant for all thicknesses and forefoot cushioning conditions, and the hard-soft pair comparison was significant for all but the short, soft forefoot condition. The same increasing (though not significant) then decreasing trend was present for all comparisons. For clarity, each curve was plotted as difference from the hard condition. For reference, the light blue curve is the same data plotted in Figure 23.

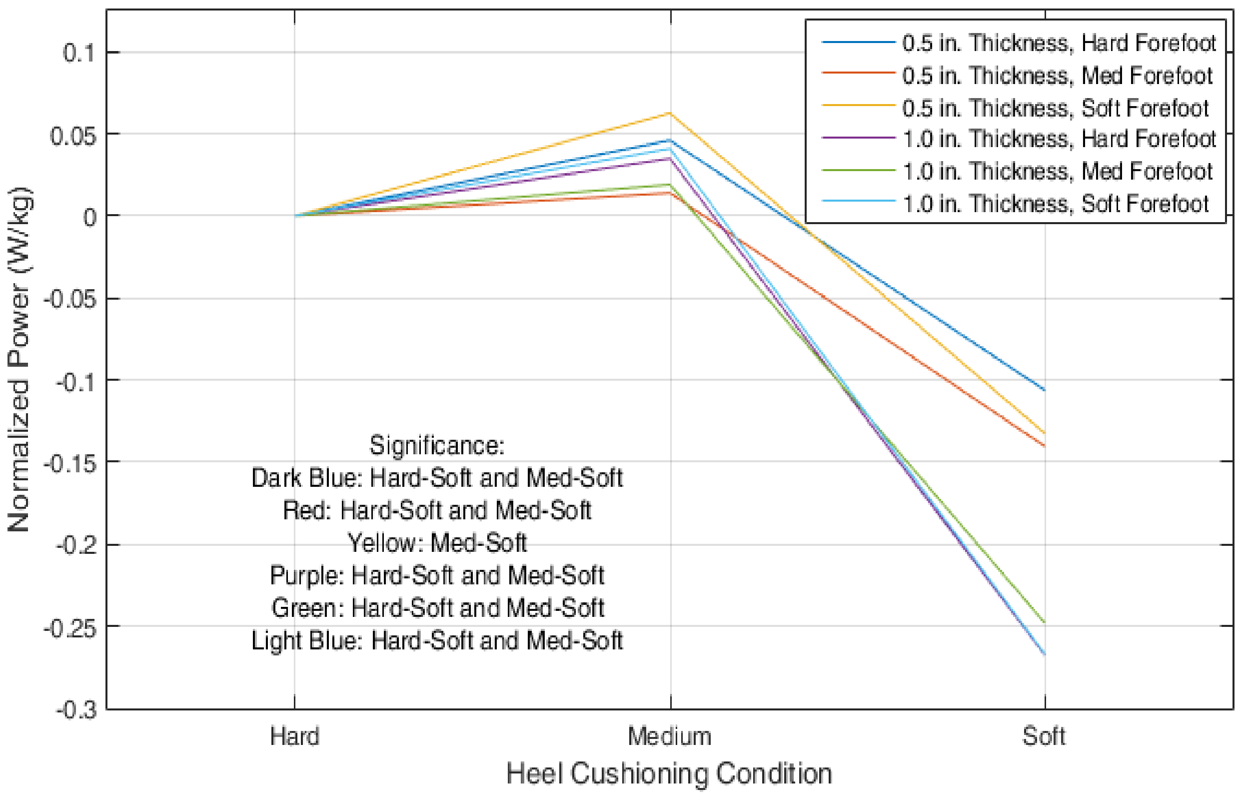


Figure 25: COM collision peak power for all combinations of forefoot cushioning and thicknesses as differences from hard heel condition. The trends found when moving towards softer heel cushioning are similar for all curves shown. Statistically significant pairs are shown in the text at the bottom-left of the Figure. Standard deviation bars were omitted for clarity

Changing heel cushioning showed little effect on COM peak push-off power, as would be expected. A representative set can be seen in Figure 26. Though analysis of this comparison yielded no statistical significance, significance was found for the hard forefoot condition and medium forefoot condition for the tall trials. That being said, the comparison with the maximum change across heel cushioning conditions was the tall, medium forefoot comparison, and the total change there was only 0.12 W/kg, less than the minimum change for the data plotted in Figure 15. Also, the trend followed by that comparison did not occur across all thicknesses and forefoot cushioning conditions.

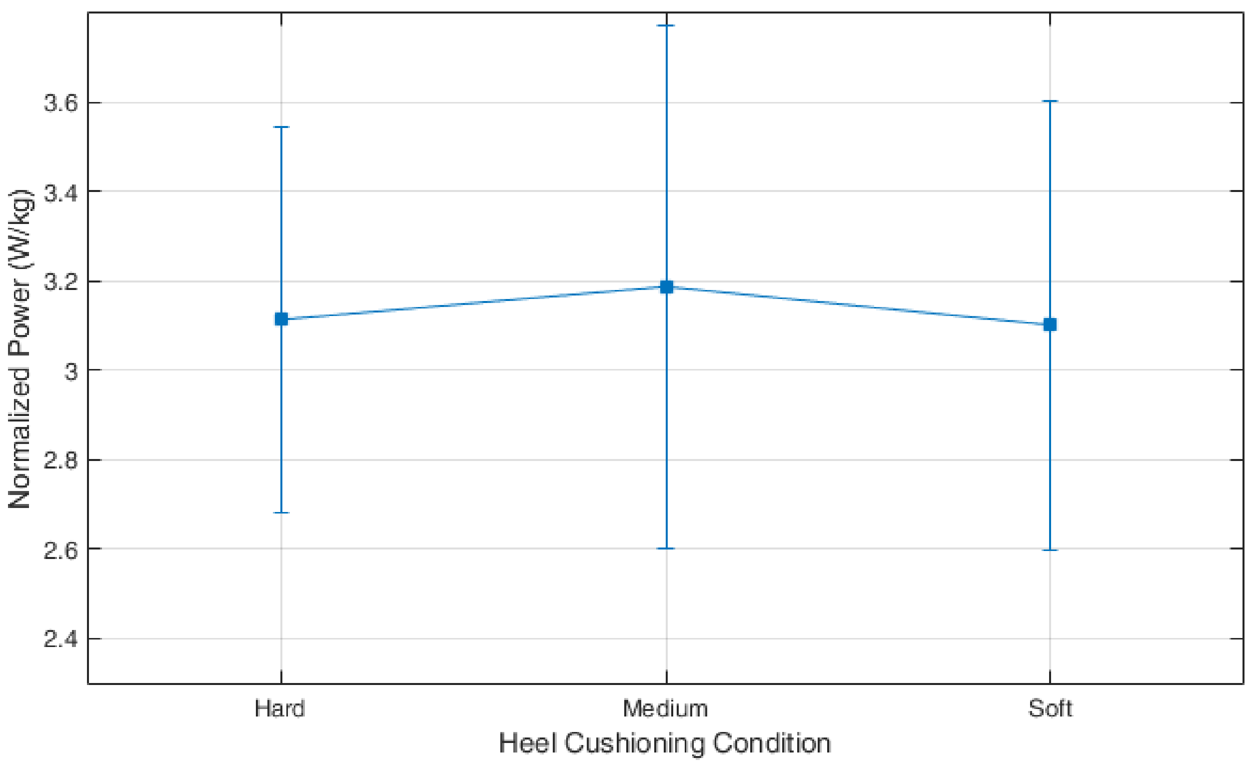


Figure 26: COM push-off peak power for hard, medium and soft heel cushioning conditions. This comparison showed no statistical significance (P = 0.28). The forefoot cushioning for this data was soft and the thickness was 25.4 mm. Most forefoot cushioning conditions showed a similar lack of meaningful trend, with only two exceptions.

Collision can also be seen well in the 3DOF+F curve. Figure 27 shows 3DOF+F power plotted over typical COM power, represented across a full stride. There is a clear power absorption phase which aligns in time with COM collision (peak occurring right around 10% stride). It is interesting to see, based on this similarity between these curves at collision, if varying the heel cushioning between hard, medium, and soft will elicit a similar effect on 3DOF+F power as did doing so for COM power.

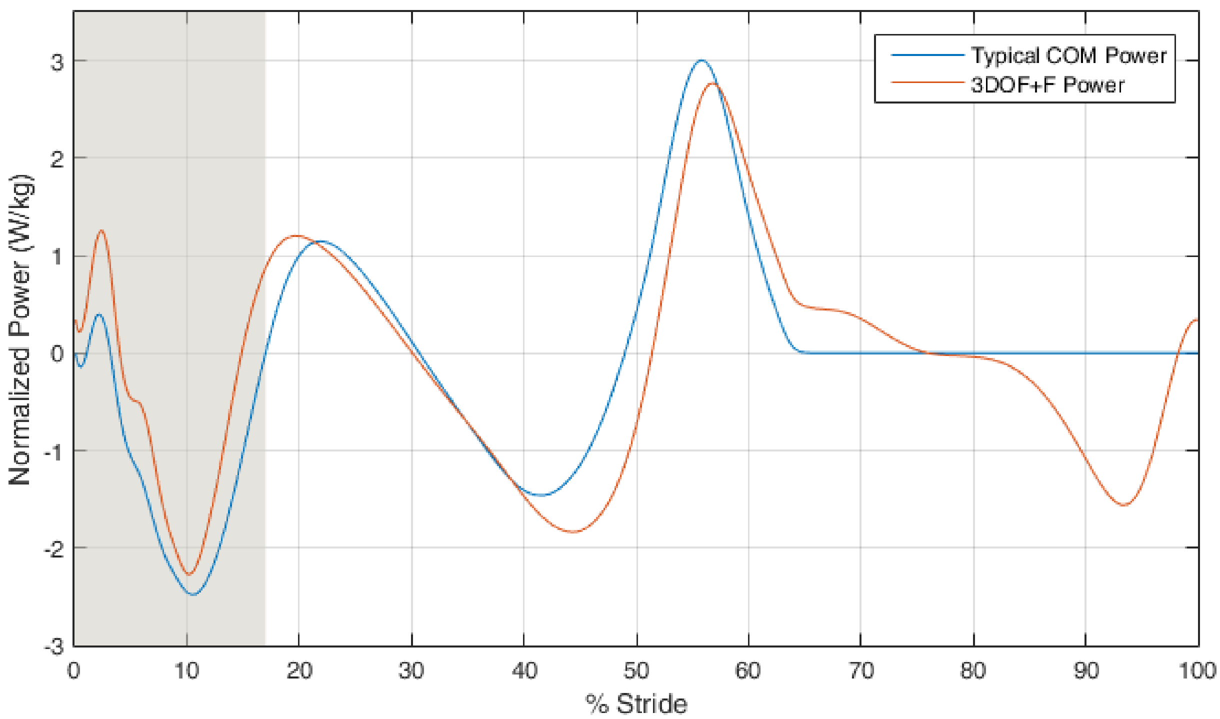


Figure 27: 3DOF+F power curve plotted over typical COM power with collision phase highlighted.

One subject's hard, medium, and soft heel 3DOF+F curves did show a similar trend to the corresponding COM power curve. Figure 28 presents the plots for this subject across a stride. Note the tendency at collision: again, the soft heel condition data has the most negative peak, followed by the hard heel condition and the medium heel condition.

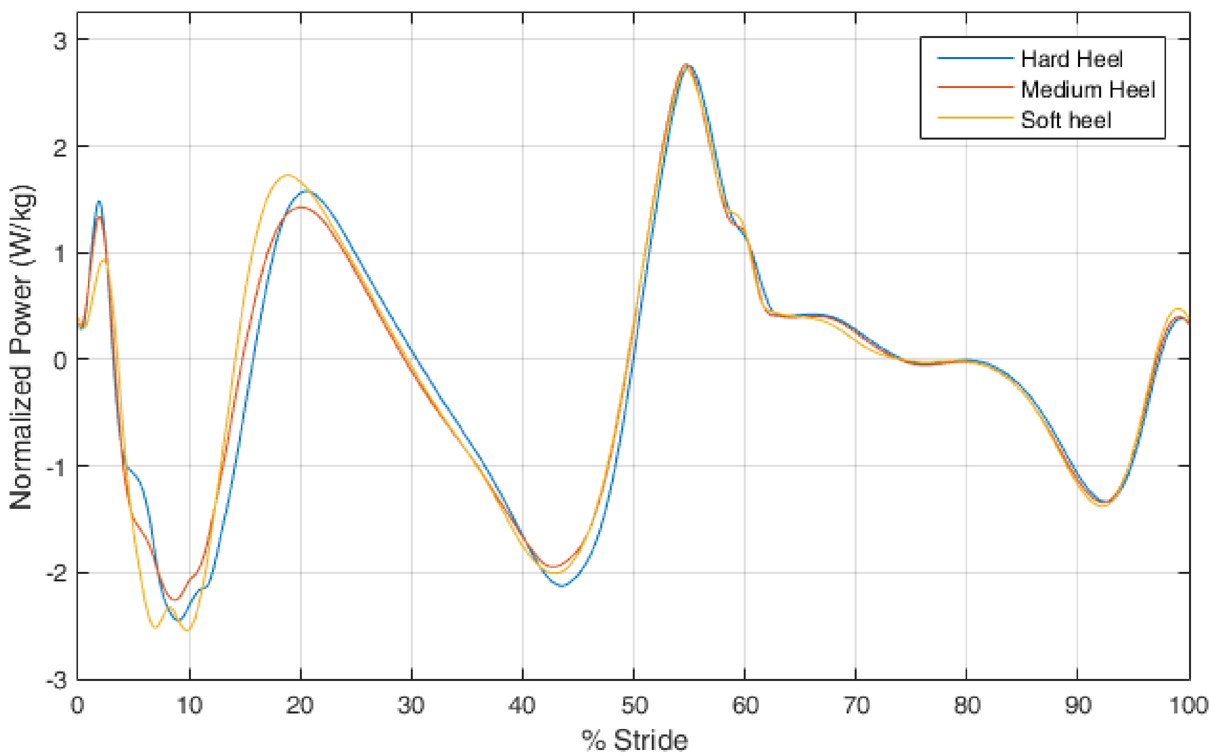


Figure 28: 3DOF+F power plots for a representative subject varying only heel cushioning. For these plots, the thickness was tall, and the forefoot cushioning was kept as soft. The waveform around the collision peak for the soft heel condition was a common result found for both thickness for a number of subjects.

Representative trends for individual contributions of the hip, knee, ankle, and foot to the 3DOF+F power are plotted in Figure 29. The values making up each mean and standard deviation were found by probing the power curves for the hip, knee, ankle, and foot at the time index corresponding with COM peak collision power. For clarity, only one condition for each joint was included. For these comparisons, the thickness seemed to play a role in how each joint was contributing power. Figure 30 shows representative ankle and knee plots for the tall trials. Interestingly, where the ankle and knee show no meaningful or significant trends in the short trials, for the tall trials they begin to show significant decreasing and increasing trends respectively.

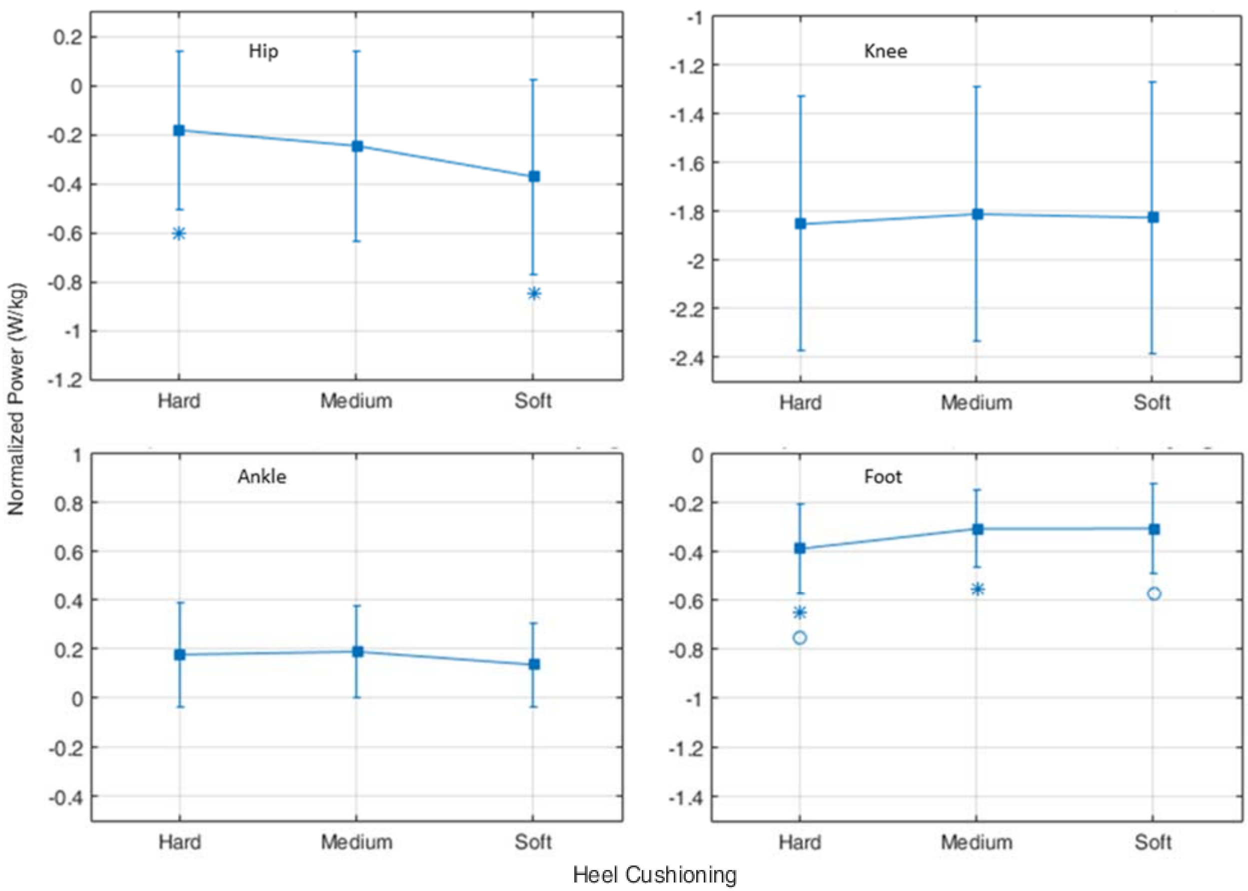


Figure 29: Contributions to 3DOF+F power at COM peak collision power. The plots shown are representative of the trends seen across heel and thickness conditions. Clockwise from the top-left, the hip (short thickness, hard forefoot) showed a significant decreasing trend, the knee (short thickness, hard forefoot) showed no meaningful trend, the foot (short thickness, medium forefoot) showed a significant increase for two comparisons, the ankle (short thickness, medium forefoot) showed no meaningful trend. The bars indicate standard deviation, and significantly different pairs are shown with symbols '*' and 'o'.

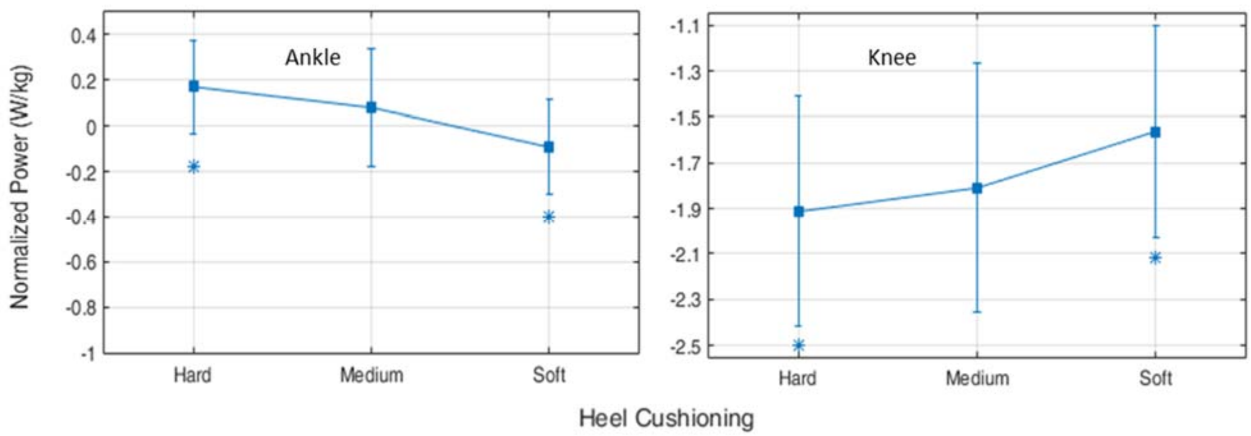


Figure 30: Ankle and knee contributions to 3DOF+F Power at COM peak collision power for thick trials. Significant pairs are designated by the symbol '*'.

Generally, the majority of subjects followed the presented trends (Figure 25) and only a small number deviated from the general shape of the curve. Figure 31 shows a plot of all subjects for the tall, hard forefoot comparison while varying heel cushioning. For clarity, the data was presented as difference from the hard condition to emphasize the similarity of intra-subject trends. For reference, this data is averaged across subjects and represented as the purple line in Figure 25.

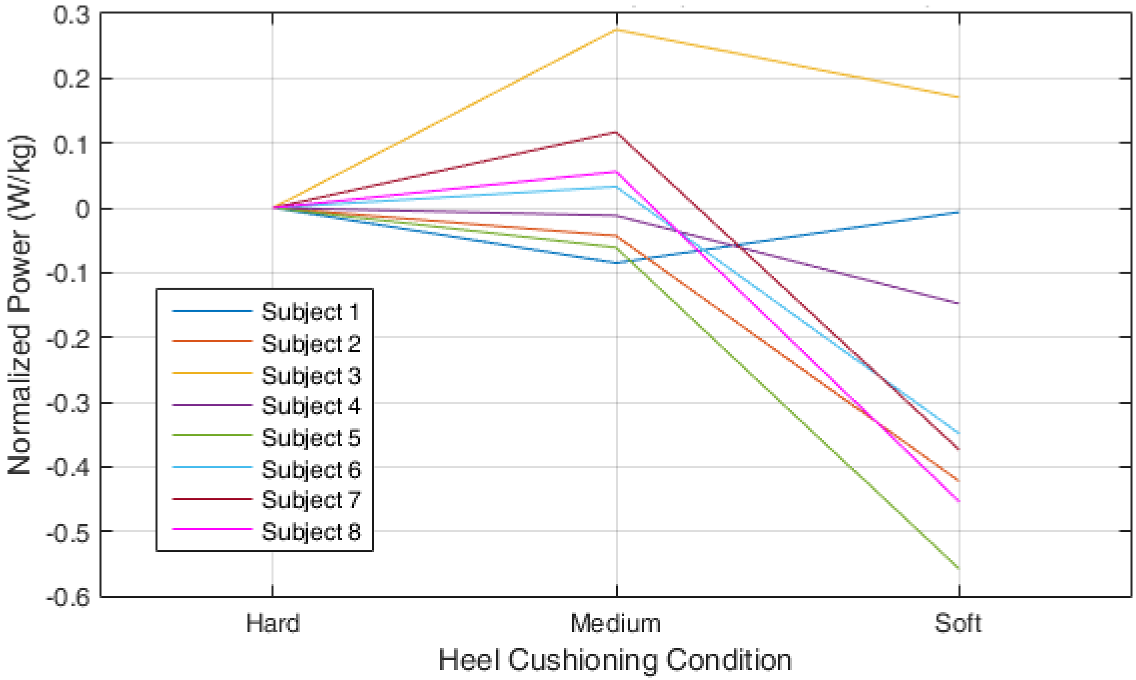


Figure 31: Hard-medium-soft heel, peak COM collision power comparison for all subjects. Note the general trend (refer to Figure 24), and the inter-subject variability. This comparison was made for the tall, hard forefoot trials across subjects.

3.3 Push-Off and Collision Work

Integrating the COM power curve over push-off and collision yield the work done on the COM by each individual leg. For push-off work, no significant differences in across-subject averages were detected as forefoot cushioning was varied for all heel cushioning and thickness conditions. Varying heel cushioning showed no recognizable general trend, though one of the curves (short, hard forefoot) showed statistical significance for the hard-soft (P = 0.010) and hard-medium (P = 0.017) comparisons. The percent increase for this curve from the hard heel to the soft heel was only 3.5%. See Figure 32 for the corresponding data plots.

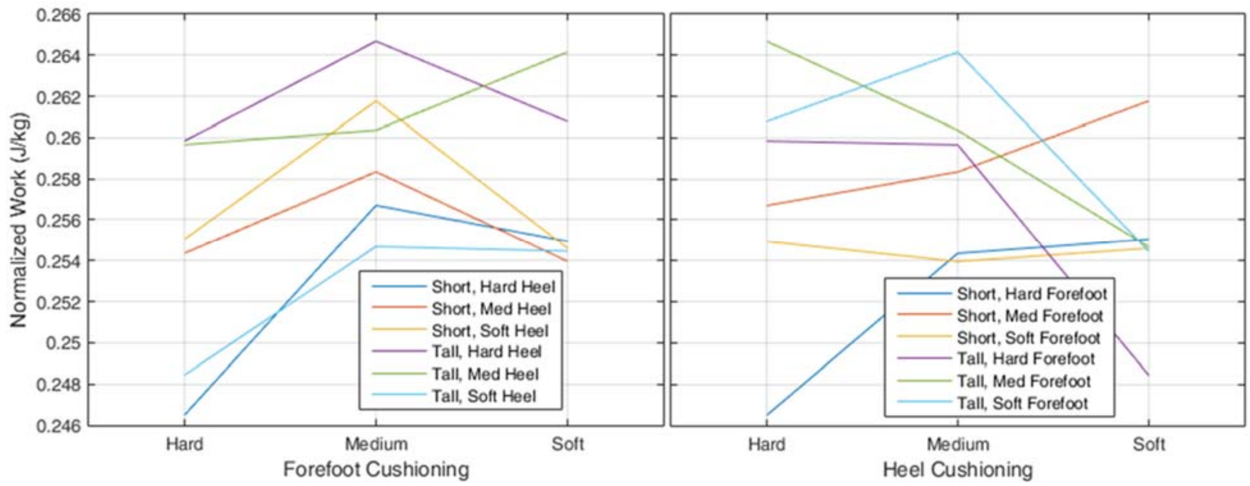


Figure 32: Push-off work comparisons varying forefoot cushioning and heel cushioning. The only significant comparison was found for the short, hard forefoot comparison while varying heel cushioning (right graph, dark blue line).

Looking into the work done by the joints and foot during push-off revealed that the shoe-foot complex absorbed less negative work as the forefoot cushioning varied from hard to soft. This calculation was performed by finding the range corresponding to push-off using the COM power plot, and integrating the deformable foot power curve over this range. When a similar approach was applied to the other joints, no substantial or significant trends were found. Figure 33 presents the foot push-off work across all thicknesses and heel conditions. The trends found seem to be grouped by thickness: for the short conditions, the comparison that had significance (hard heel, hard-soft forefoot, P = 0.014, dark blue curve in Figure 33) showed change in work of 0.0163 J/kg, corresponding to 13.3% less negative work for the soft forefoot condition. For the tall conditions, a sample comparison (hard heel, hard-soft forefoot, purple curve in Figure 33) showed change in work of 0.0381 J/kg, corresponding to 31.9% less negative work for the soft forefoot condition.

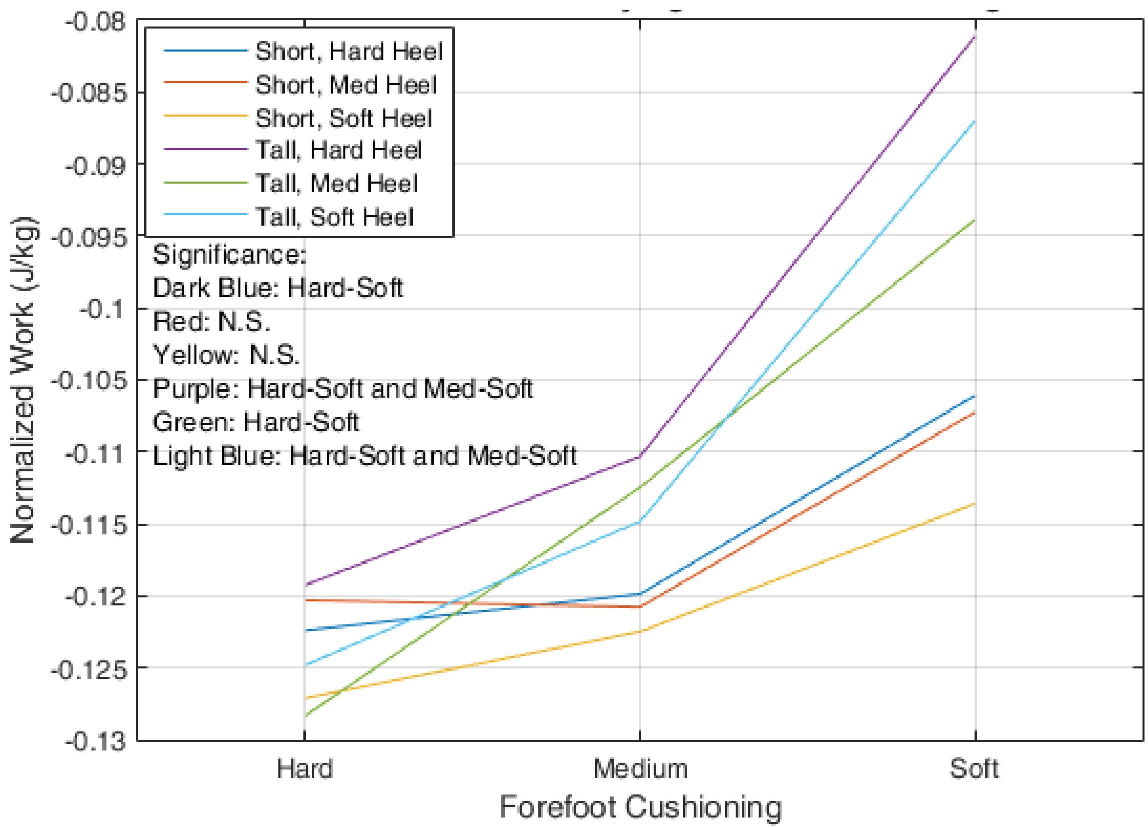


Figure 33: Deformable foot push-off work varying forefoot cushioning. For each curve, the thickness and heel cushioning are constant. Interestingly, the results seem to be grouped by thickness. Significant comparisons are presented in the text on the left.

Interestingly, for collision work, significant differences and apparent trends were found both when varying forefoot cushioning and when varying heel cushioning. Figure 34 shows the collision work response from varying forefoot cushioning. Four of the six curves plotted show significance in the hard-soft forefoot comparison, and two of those four curves also show significance in the medium-soft comparison. For a representative curve (tall, medium heel, green curve in Figure 34), the maximum change from hard to soft forefoot cushioning was -0.0174 J/kg, corresponding to value 7.8% more negative.

Figure 35 shows the collision work response from varying heel cushioning. Two of the six curves plotted show statistical significance in the hard-soft heel comparison, and one of those curves also shows significance for the other two comparisons (hard-med and med-soft). For a representative curve (tall, hard forefoot, purple curve in Figure 35), the maximum change from hard to soft heel cushioning was -0.0124 J/kg, corresponding to a value 5.7% more negative.

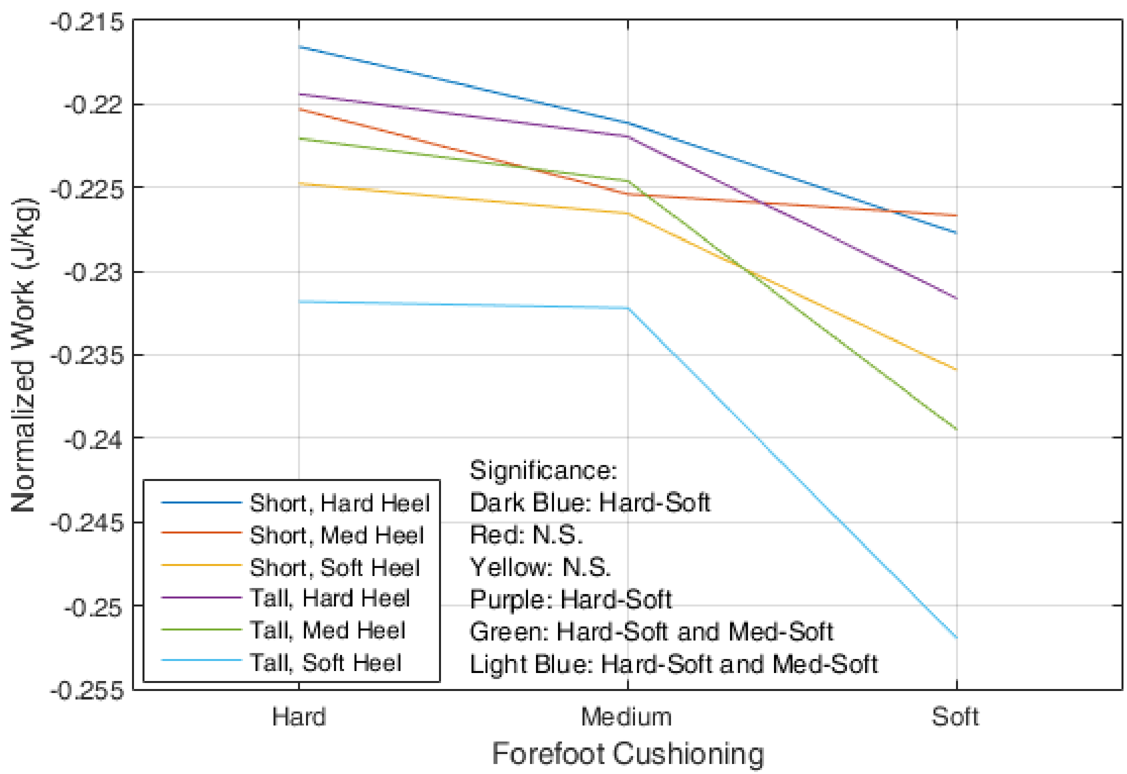


Figure 34: COM collision work varying forefoot cushioning. For each curve, the thickness and heel condition remained constant. Significant comparisons are displayed in the text at the bottom of the figure.

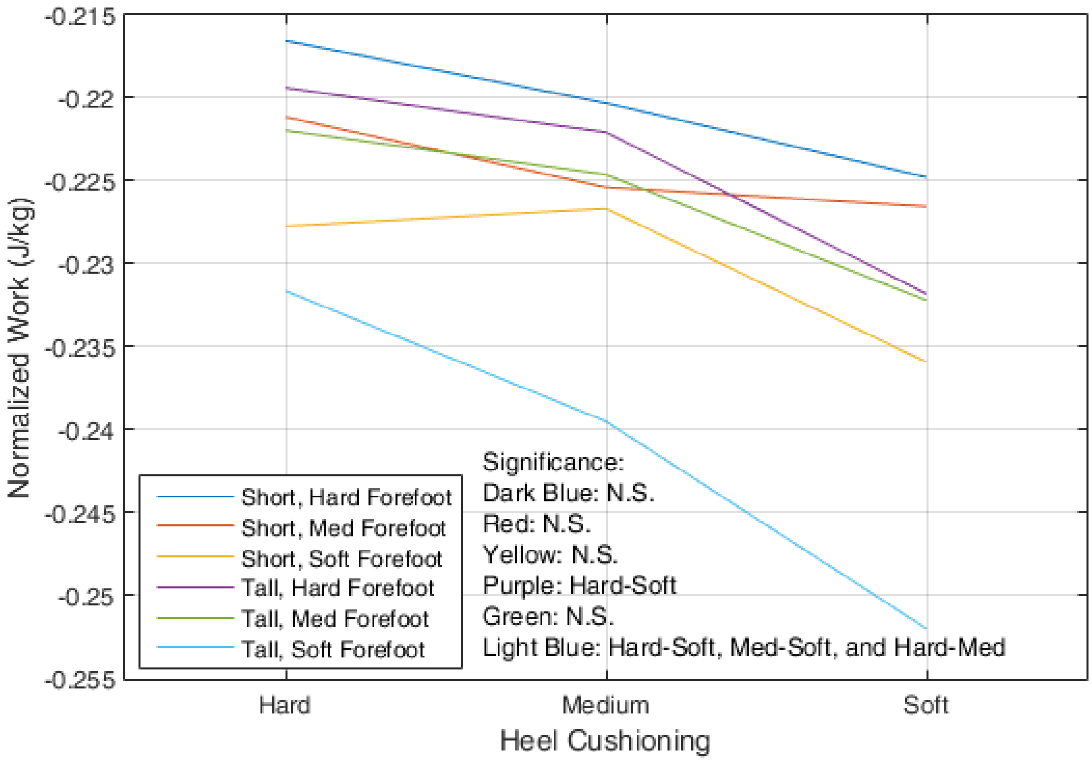


Figure 35: COM collision work varying heel cushioning. For each curve, the thickness and forefoot condition remained constant. Significant comparisons are displayed in the text at the bottom of the figure.

Looking into the work done by the joints and foot at collision revealed that the foot-shoe complex again was the only entity making up the 3DOF+F curve that showed significant work trends, this time becoming more negative as heel cushioning was varied from hard to soft. Though it appears a consistent trend across all curves in Figure 36 that the work becomes less negative from hard to medium, then more negative from medium to soft, no significance was determined between the hard heel and medium heel conditions for any of the curves. Interestingly, only the tall conditions showed significance. For an example comparison (tall, medium forefoot, medium-soft heel, green curve in Figure 36), the maximum change from the medium to soft heel cushioning was -0.0099 J/kg, corresponding to a value 93% more negative.

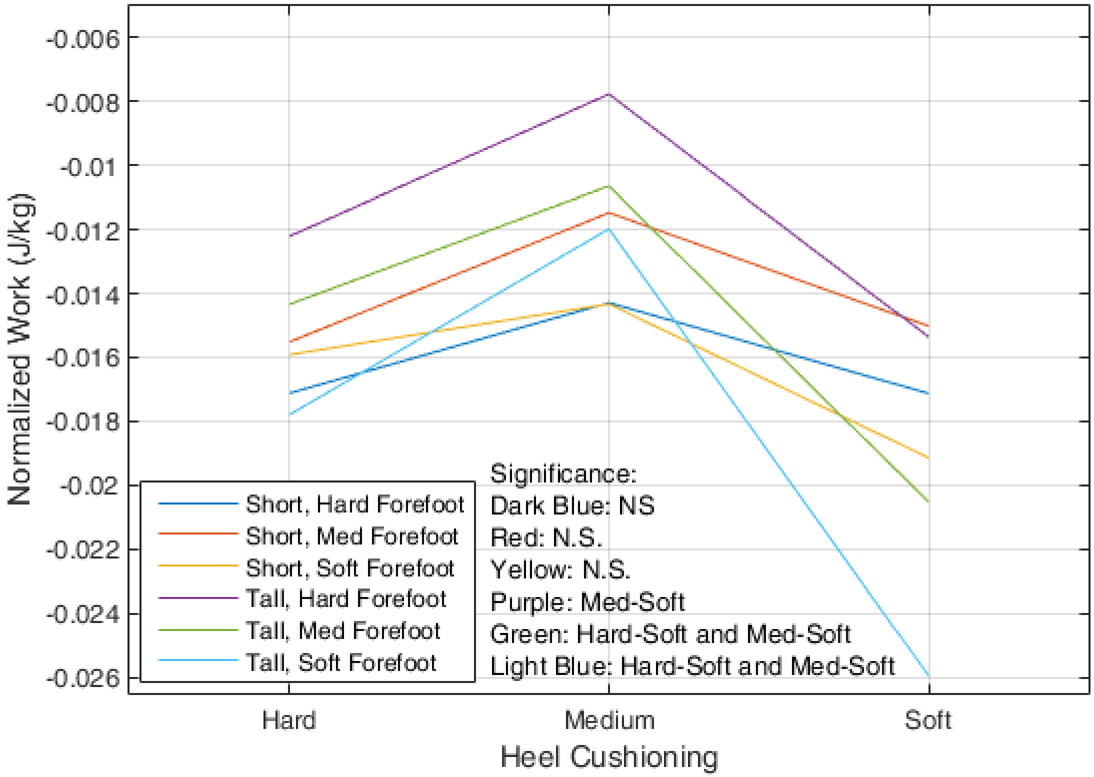


Figure 36: Deformable foot collision work varying heel cushioning. For each curve, the thickness and forefoot cushioning are constant. Significant comparisons are presented in the text at the bottom of the plot. Only tall trials showed significant comparisons.

3.4 Stiff Forefoot Comparisons

The comparison between the hard and stiff forefoot cushioning conditions (short with a constant hard heel) produced interesting results. Table 4 below summarizes these results. The COM push-off peak power decreased by 0.15 W/kg when the hard forefoot was swapped with the stiff forefoot, corresponding to a percent decrease of 5.1%. The COM collision peak power did not change significantly. Neither the push-off nor the collision work changed substantially or significantly. Measurements of the hip, knee, ankle, and foot power at the COM push-off power peak index showed that the knee and foot power were not affected by the change in forefoot cushioning, but the ankle and hip both had significant ($P < 0.05$) changes. The ankle power decreased by 0.23 W/kg and the hip power increased by 0.03 W/kg. For simplicity of presentation, only the significant trends in joint and foot work for push-off and collision are shown. The hip push-off work increased by 0.005 J/kg.

Table 4: Hard-Stiff Forefoot Comparisons with a Short, Hard Heel

	Hard Forefoot	**Stiff Forefoot**	**T-Test Significance**
COM Push-off Peak Power	2.97 W/kg	2.82 W/kg	P = 0.029
COM Collision Peak Power	-2.52 W/kg	-2.55 W/kg	P = 0.59
Hip Push-off Power	0.88 W/kg	0.91 W/kg	P = 0.035
Knee Push-off Power	-0.51 W/kg	-0.52 W/kg	P = 0.90
Ankle Push-off Power	3.39 W/kg	3.16 W/kg	P = 0.010
Foot Push-off Power	-0.41 W/kg	-0.40 W/kg	P = 0.50
COM Push-off Work	0.246 J/kg	0.249 J/kg	P = 0.79
COM Collision Work	-0.217 J/kg	-0.217 J/kg	P = 0.95
Hip Push-off Work	0.147 J/kg	0.152 J/kg	P = 0.003

3.5 Step Frequency

When averaged across subjects, step frequency showed no significant changes for hard-medium-soft forefoot comparisons and for hard-medium-soft heel comparisons for both thicknesses, short and tall. For these comparisons, the maximum change in step frequency for the subject averages was 1.2 steps/min, and subject-specific results never yielded a change in step frequency above 3.3 steps/min. For the comparisons between stiff and hard forefoot conditions, the average step frequency change across subjects was less than 0.1 steps/min, but the maximum subject-specific change in step frequency was almost 4.0 steps/min.

CHAPTER 4

DISCUSSION

Empirical evidence was observed:

1. supporting Hypothesis One: forefoot and heel cushioning effects can largely be decoupled during walking.
2. refuting Hypothesis Two: softer forefoot cushioning actually led to increased COM peak push-off power.
3. supporting Hypothesis Three: softer heel cushioning led to more negative COM peak collision power.

4.1 Forefoot and Heel Cushioning Effects Can Be Decoupled

Hypothesis one was supported by the trends, or more accurately, lack of trends found for COM peak collision power and peak push-off power for varying forefoot and heel respectively. Not only were the comparisons not statistically significant, the difference magnitudes were also not substantial compared to the magnitude of differences observed for COM push-off power and COM collision power varying forefoot and heel, respectively (refer to Figure 16 and Figure 26). We therefore concluded that forefoot and heel cushioning effects could be largely decoupled. Thus, for Hypothesis Two we only assessed the effect of forefoot on push-off, and for Hypothesis Three we only assessed the effect of heel on collision.

Further evidence in support of Hypothesis One can be found in the lack of trends found in inverse-dynamics-estimated joint powers. For all short trials, there were no significant trends found for any joint power or for foot power when comparing push-off across heel cushioning conditions or when comparing collision across forefoot cushioning conditions. The same can be said about the tall trials, with the one exception of a solitary statistically significant comparison in knee collision power (medium-soft forefoot comparison, medium heel).

The results derived from COM power support Hypothesis One. A secondary measure, COM collision work, shows changes coinciding with changing forefoot cushioning (Figure 34). Since COM power is the primary analysis metric and joint and foot power results support the findings in COM power, though the COM collision work results are interesting, they do not directly refute Hypothesis One.

The data presented in this work supports the idea that forefoot and heel cushioning effects can be largely decoupled when considering their effects on peak collision and push-off power. This concept can be very useful in the design of purposeful walking shoes. If it is determined that a specific individual could benefit from a shoe that decreases COM push-off power, it may be possible to design a shoe with custom sole properties at the forefoot to satisfy that need without adverse effects at collision.

4.2 Softer Forefoot Cushioning Increases COM Peak Push-Off Power

Hypothesis two was not supported by the trends presented by this work. In fact, COM push-off peak power increased as the forefoot cushioning was made softer (refer to Figure 15).

Based on the deformable foot power estimates presented, the foot-shoe combination did not seem to absorb more power (at peak push-off) when the forefoot cushioning was softer (see Figure 19). This may be because the majority of absorption in the forefoot sole happens during the rebound and pre-load phases of walking, when that section of the shoe is initially compressed. As a result the forefoot cushioning has a less substantial effect on push-off peak power. Figure 37 supports this explanation. As the forefoot cushioning becomes softer, the power absorption occurring before 50% stride (beginning of push-off), and qualitatively the absorptive work during the pre-push-off phase, becomes more negative.

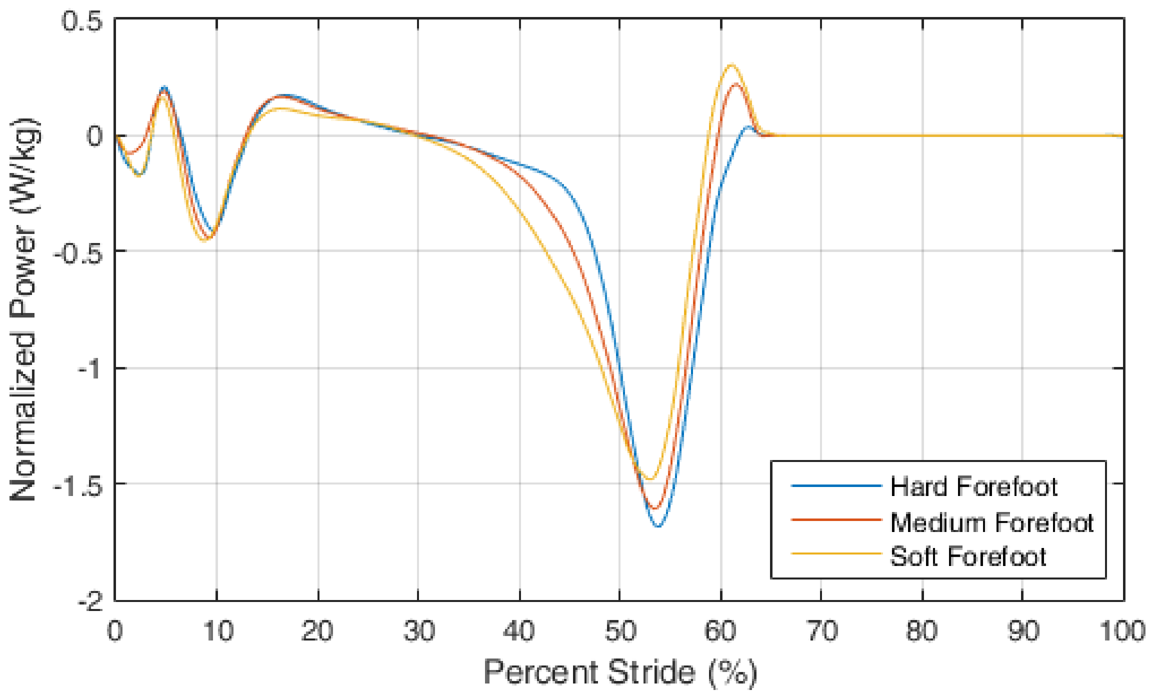


Figure 37: Deformable foot power for a representative subject for hard, medium, and soft forefoot conditions. This comparison was made for the tall condition with a hard heel. Average power and work occurring during the pre-load phase of walking gait qualitatively seemed to become more negative, potentially explaining why no trends were found for foot power at push-off. The peak trends visible near 53% stride were not found across subject averages.

Inverse dynamics allow a glimpse into what the biological system is contributing to make up these trends. The net power generated by the ankle musculature, at least for the short conditions, increases with a comparable magnitude to the increases noted in COM power (see Figure 19). This implies that the average subject adapts to softer forefoot cushioning by increasing ankle power, and this adaption manifests as increased push-off power to the COM.

Since the magnitude of mechanical work done during push-off was not found to change with changing forefoot cushioning (see Figure 32), a change in peak COM push-off power can be used as a measure of impulsivity of that push-off work delivery. This suggests that softer forefoot cushioning tends to hasten the push-off process, making it more impulsive. Interestingly, foot-shoe complex work done at push-off became less negative with softer forefoot cushioning for all heel and thickness conditions (see Figure 33).

None of the joint works about the ankle, knee, and hip showed substantial trends due to varying forefoot cushioning. COM and joint push-off works did not change but foot-shoe work became less negative, so conservation of energy informs that this missing work must have been accounted for somewhere. For the presented trend in Figure 33, the change of 0.0381 J/kg, using an average subject mass of 77.9 kg, corresponds to 3.97 J of push-off work. A typical push-off work for the average subject (for the tall, hard forefoot, hard heel condition) was 20.2 J. This leaves a change in work corresponding to 20% of total push-off work. There are many ways this work may be missing from the biomechanical analysis presented. A more rigorous analysis of joint powers (i.e., 6DOF analysis, Zelik et al. 2015), the inclusion of peripheral power terms (Zelik et al. 2015), or analysis of the upper body may be needed to account for this missing energy.

4.3 Softer Heel Cushioning Leads to a More Negative COM Peak Collision Power

Hypothesis three was supported by the trends presented by this work. It was found that, under constant thickness and forefoot cushioning, decreasing the heel cushioning durometer from hard to medium had insignificant effect on COM peak collision power, but further decreasing to the soft durometer resulted in a substantial and significant more negative COM peak collision power, as predicted (see Figure 24). That hard and medium heel cushioning did not have a significant effect on power was not predicted, but an explanation may be that though the heel durometer varies roughly linearly (see Section 2.2), its effect on collision does not, and there is potentially a heel cushioning durometer that results in a COM peak collision power that is the least negative.

The decreasing trend in collision work under the same comparison conditions (see Figure 35) further supports the claim that the heel portion of the sole is actively absorbing power during the collision phase, dissipating energy on top of the energy being dissipated by the body itself. Since the collision work generally became more negative as heel cushioning varied from hard to medium to soft, the hard-medium comparison resulted in decreased work but not in decreased power. This implies that the medium heel condition caused the work to be delivered to the COM in a less impulsive manner when compared to the hard heel. Since impulsive forces on the musculoskeletal system have been associated with increased risk of tissue damage and other injuries (Collins and Whittle 1989, Radin et al. 1973, Nigg et al. 1987, Nigg 1997, Lafortune and Hennig 1992, Voloshin and Wosk 1981), this effect can potentially be interpreted as beneficial. For the soft heel, COM collision power does decrease significantly compared to the hard and medium heel, and the collision work does as well, though to a proportionally lesser extent (12.4%

more negative for power, 5.7% more negative for work). This implies that though the work becomes more negative, it likely also becomes more impulsive, with higher magnitude peak powers. A possible explanation of this is that the soft heel cushioning was so soft, its compression absorbed little energy before the heel “bottomed out”, resulting in a more impulsive peak power. From this data, it can be surmised that heel cushioning of a medium durometer may be comparatively beneficial for decreasing impulsivity of collision loadings.

Joint-level measurements (see Figure 29) did not support the corollary to this hypothesis: that the body would provide less negative collision power due to that absorbed by the foot. Hip followed a trend of more negative power when heel cushioning varied from hard to soft which was of similar magnitude to the overall decrease from hard to soft for COM power. The effect noted above for the medium heel condition cannot be supported or refuted by hip power metrics since the medium heel condition effect on hip power was not significantly different from the effect of either the hard or the soft condition.

Deformable foot power measurements (see Figure 29) interestingly showed a trend of less negative collision power for medium and soft heels compared to the hard heel. The hard-soft comparison supports the explanation made above that the soft condition was too soft to absorb power at peak collision. That the medium heel shows less negative power than the hard heel may be a partial explanation for why the COM peak collision power does not vary significantly in the same comparison, especially given that hip power seems to be more negative (though not significantly) for that comparison.

Foot work during collision followed a similar decreasing trend for the hard-soft and medium-soft heel comparisons (see Figure 36), though only significant for the tall conditions.

That the hard-medium comparison showed no statistically significant differences but the COM collision work generally became more negative (see Figure 35) from hard to medium to soft heel conditions signifies that some negative work done to the COM for the medium heel condition is potentially not being accounted for in the metrics utilized in this thesis (possibly due to biomechanical measurement limitations or unmeasured soft tissue dynamics elsewhere in the body). The differences in this case are small enough to be reasonably assumed as error in measurement, and more importantly, these relatively small discrepancies are completely independent of the ability to evaluate the key main hypotheses in this study.

4.4 The Decreased Impulsivity of Push-Off with Higher Bending Stiffness

When the hard forefoot was swapped for the stiff forefoot (see Table 4), a number of key metrics were altered that point to the decreased impulsivity of push-off with higher bending stiffness. Though the COM push-off work was not statistically different for the stiff condition, the push-off peak power decreased significantly, implying that the stiffer (in bending) forefoot condition caused the same push-off work to be delivered in a less impulsive way. This result may point to the role of the toe joint in walking (Hicks 1954, Sarrafian 1987), since the stiffness of the forefoot acts in parallel to the stretching of the plantar aponeurosis.

4.5 Limitations

One challenge in the data analysis for this experiment is the varying shape of COM power during collision. As noted in Figure 28, two distinct peaks were often observed for soft heel conditions. Thus quantifying peak power values only provided a partial summary of the power waveform differences.

Utilizing inverse dynamics to estimate joint-level power has error sensitivity issues. Zelik and Kuo (2010) discuss, for example, that perturbing the location of the knee joint center has a substantial effect on the overall shape of the knee power curve during double support. Since markers were attached to all subjects in a similar way, the model and technique used to calculate inverse dynamics was the same across subjects. Thus analysis was focused solely on relative differences, which helped mitigate these common accuracy issues.

There is some evidence in this experiment that the tall, soft forefoot condition reduced the stability of walking. Observing the trends found for COM push-off power in all trials (see Figure 15) and ankle push-off power (see Figure 19) in short trials, one would assume that ankle power would increase at the same rate as COM power for the tall trials as well. That this was not observed (see Figure 20) was potentially due to instability. Without being prompted, multiple subjects commented that it was difficult to walk on that condition, or that it was difficult for them to maintain balance. One subject was visibly off-balance during a substantial portion of that trial. Those comments were not made and balance issues were not noted for the short, soft forefoot condition or for any of the other tall conditions. This leads one to believe that some of the trends associated with the tall, soft forefoot (namely ankle power) may be considerably affected by more indirect factors not present to the same extent for the other conditions.

A final limitation is related to the subject group tested. Due to the shoes used in this study, the participants were all males of similar foot size. All subjects tested were young and healthy.

From these data it is unclear if the findings are generalizable to females, or more elderly populations.

4.6 Statistical Power Analysis

A power analysis was conducted on the presented trends (Figure 14 and Figure 24). It was determined that for the push-off comparison (Figure 14), given the number of subjects, the statistical power of the hard-soft comparison was 0.93. When the statistical power was calculated for this comparison given that an ANOVA was used for the original significant determination, the power yielded was reduced to 0.66. Similar calculation for the collision trend (Figure 24) yielded a power value of 0.96 and 0.76 for the medium-soft comparison. This signifies that based on the experimental design (i.e., number of subjects included), it can be said that the overall trends in the data can be trusted with confidence, but experiment was slightly too underpowered to make an equally confident claim about the nuances of the hard-medium-soft comparisons. Preliminary calculations yielded that the addition of six subjects (for a total subject pool of N = 14), would bring all power values to above 0.9 for three-way ANOVA comparisons.

CHAPTER 5

CONCLUSION

5.1 Impact

The impact of this thesis is to begin the work towards a more comprehensive understanding of the effect of sole cushioning on biomechanical aspects of walking. This understanding could potentially be utilized to tailor footwear to alter/enhance aspects of healthy walking or for gait rehabilitation. Based on the results of the presented experiment, the forefoot and heel cushioning parameters may be independently altered to have some desired effect on the biomechanics of healthy gait. Manipulating forefoot cushioning primarily influenced aspects of push-off and manipulating heel cushioning primarily influenced aspects of collision.

There is a demonstrated interest in building running shoes with customized sole properties utilizing additive manufacturing (New Balance 2015), and this technology can easily be applied to building equally as customized walking shoes (or shoes for other specific activities). A shoe with a midsole tuned to decrease impulsivity of push-off and collision, for example, might be fabricated to have a moderately soft heel, but a stiffer, harder forefoot. Alternatively, if an individual is found to have certain gait asymmetries, say that the right leg does not provide as much peak power during push-off, a pair of rehabilitation shoes may be fabricated such that the right shoe has softer a forefoot midsole to address the asymmetry.

5.2 Future Work

An interesting next step in the understanding of midsole cushioning effects would be to design an experiment to determine the metabolic effects of forefoot and heel cushioning. The results of this thesis would inform condition selection to identify a smaller number of conditions for which to measure metabolic expenditure. The results of this metabolic analysis could then inform the construction of a shoe midsole for the purpose of increasing gait economy.

Another interesting experiment could be to repeat the analysis completed in this work (and/or run a metabolic analysis) but for varying midsole elasticity or viscoelasticity. Since elastic return may be an important factor in walking, it is possible that materials that return more energy than they dissipate could have an important place in the customized tailoring of shoe midsoles for many different purposes. A future work could also benefit from a stability analysis, since midsoles which result in unstable gait should likely be excluded from further analysis.

Furthermore, a future experiment could benefit from including more rigorous analysis of total power from the joints (6DOF power estimates) and the peripheral power to accelerate the lower extremities during gait (Zelik et al. 2015). This analysis may provide further insights into the interplay between contact-surface dynamics and energy absorbed in the biological system.

Finally, it would be interesting in a future work to more finely resolve the effects of cushioning by including conditions between those tested in this thesis. The effects of materials with shore A durometer readings of 10, 30, and 50 could help turn the trends discovered in this thesis into more defined curves.

REFERENCES

Adamczyk, Peter G., and Arthur D. Kuo. 2009. “Redirection of Center-of-Mass Velocity during the Step-to-Step Transition of Human Walking.” Journal of Experimental Biology 212 (16): 2668–78. doi:10.1242/jeb.027581.

Arampatzis, Adamantios, Gert-Peter Brüggemann, and Verena Metzler. 1999. “The Effect of Speed on Leg Stiffness and Joint Kinetics in Human Running.” Journal of Biomechanics 32 (12): 1349–53. doi:10.1016/S0021-9290(99)00133-5.

Boyer, Katherine A., and Benno M. Nigg. 2004. “Muscle Activity in the Leg Is Tuned in Response to Impact Force Characteristics.” Journal of Biomechanics 37 (10): 1583–88. doi:10.1016/j.jbiomech.2004.01.002.

Brockway, JM. 1987. “Derivation of Formulae Used to Calculate Energy Expenditure in Man.” Human Nutrition. Clinical Nutrition 41 (6): 463–71.

Burkett, LN, WM Kohrt, and R Buchbinder. 1985. “Effects of Shoes and Foot Orthotics on VO2 and Selected Frontal Plane Knee Kinematics.” Medicine and Science in Sports and Exercise 17 (1): 158–63.

Burton, Alan C. 1935. “Human Calorimetry. 2. The Average Temperature of the Tissues of the Body.” Journal of Nutrition 9: 261–80.

Cappozzo, A., A. Cappello, U.D. Croce, and F. Pensalfini. 1997. “Surface-Marker Cluster Design Criteria for 3-D Bone Movement Reconstruction.” IEEE Transactions on Biomedical Engineering 44 (12): 1165–74. doi:10.1109/10.649988.

Cappozzo, A., F. Figura, M. Marchetti, and A. Pedotti. 1976. “The Interplay of Muscular and External Forces in Human Ambulation.” Journal of Biomechanics 9 (1): 35–43. doi:10.1016/0021-9290(76)90137-8.

Cardillo G. 2008. “Anovarep: compute the Anova for repeated measures and Holm-Sidak test for multiple comparisons if Anova is positive.” http://www.mathworks.com/matlabcentral/fileexchange/18746

Cavagna, G. A. 1975. “Force Platforms as Ergometers.” Journal of Applied Physiology 39 (1): 174–79.

Cavagna, G. A., N. C. Heglund, and C. R. Taylor. 1977. “Mechanical Work in Terrestrial Locomotion: Two Basic Mechanisms for Minimizing Energy Expenditure.” American Journal of Physiology - Regulatory, Integrative and Comparative Physiology 233 (5): R243–61.

Cavagna, G A, H Thys, and A Zamboni. 1976. “The Sources of External Work in Level Walking and Running.” The Journal of Physiology 262 (3): 639–57.

Collins, J. J., and M. W. Whittle. 1989. “Impulsive Forces during Walking and Their Clinical Implications.” Clinical Biomechanics 4 (3): 179–87. doi:10.1016/0268-0033(89)90023-5.

Dauncey, M. J. 1980. “Metabolic Effects of Altering the 24 H Energy Intake in Man, Using Direct and Indirect Calorimetry.” British Journal of Nutrition 43 (02): 257–69. doi:10.1079/BJN19800089.

Dean, J. C., and A. D. Kuo. 2009. “Elastic Coupling of Limb Joints Enables Faster Bipedal Walking.” Journal of The Royal Society Interface 6 (35): 561–73. doi:10.1098/rsif.2008.0415.

Divert, C., G. Mornieux, P. Freychat, L. Baly, F. Mayer, and A. Belli. 2008. “Barefoot-Shod Running Differences : Shoe or Mass Effect?” International Journal of Sports Medicine 29 (6): 512–18.

Donelan, J. Maxwell, Rodger Kram, and Kuo Arthur D. 2001. “Mechanical and Metabolic Determinants of the Preferred Step Width in Human Walking.” Proceedings of the Royal Society of London B: Biological Sciences 268 (1480): 1985–92. doi:10.1098/rspb.2001.1761.

Donelan, J. Maxwell, Rodger Kram, and Arthur D. Kuo. 2002a. “Simultaneous Positive and Negative External Mechanical Work in Human Walking.” Journal of Biomechanics 35 (1): 117–24. doi:10.1016/S0021-9290(01)00169-5.

Donelan, J. Maxwell, Rodger Kram, and Arthur D. Kuo. 2002b. “Mechanical Work for Step-to-Step Transitions Is a Major Determinant of the Metabolic Cost of Human Walking.” Journal of Experimental Biology 205 (23): 3717–27.

Ferrannini, Eleuterio. 1988. “The Theoretical Bases of Indirect Calorimetry: A Review.” Metabolism 37 (3): 287–301. doi:10.1016/0026-0495(88)90110-2.

Ferris, Daniel P, Kailine Liang, and Claire T Farley. 1999. “Runners Adjust Leg Stiffness for Their First Step on a New Running Surface.” Journal of Biomechanics 32 (8): 787–94. doi:10.1016/S0021-9290(99)00078-0.

Ferris, D P, M Louie, and C T Farley. 1998. “Running in the Real World: Adjusting Leg Stiffness for Different Surfaces.” Proceedings of the Royal Society B: Biological Sciences 265 (1400): 989–94.

Franz, Jason R, Corbyn M Wierzbinski, and Rodger Kram. 2012. “Metabolic Cost of Running Barefoot versus Shod: Is Lighter Better.” Med Sci Sports Exerc 44 (8): 1519–25.

Frederick, E. C. 1984. “Physiological and Ergonomics Factors in Running Shoe Design.” Applied Ergonomics 15 (4): 281–87. doi:10.1016/0003-6870(84)90199-6.

Frederick, EC, TE Clarke, JL Larsen, and LB Cooper. 1983. “The Effects of Shoe Cushioning on the Oxygen Demands of Running.” Biomechanical Measurement of Running Shoe Cushioning Properties. Biomechanical Aspects of Sport Shoes and Playing Surfaces. University of Calgary, Calgary, AB, 107–14.

Gordon, D., E. Robertson, and David A. Winter. 1980. “Mechanical Energy Generation, Absorption and Transfer amongst Segments during Walking.” Journal of Biomechanics 13 (10): 845–54. doi:10.1016/0021-9290(80)90172-4.

Griffin, Timothy M., Thomas J. Roberts, and Rodger Kram. 2003. “Metabolic Cost of Generating Muscular Force in Human Walking: Insights from Load-Carrying and Speed Experiments.” Journal of Applied Physiology 95 (1): 172–83. doi:10.1152/japplphysiol.00944.2002.

Hanson, N. J., K. Berg, P. Deka, J. R. Meendering, and C. Ryan. 2011. “Oxygen Cost of Running Barefoot vs. Running Shod.” International Journal of Sports Medicine 32 (06): 401–6.

Hardin, Elizabeth C, Antonie J van den Bogert, and Joseph Hamill. 2004. “Kinematic Adaptations during Running: Effects of Footwear, Surface, and Duration.” Medicine and Science in Sports and Exercise 36 (5): 838–44.

Hicks, JH. 1954. “The Mechanics of the Foot: II. The Plantar Aponeurosis and the Arch.” Journal of Anatomy 88 (Pt 1): 25.

Kerdok, Amy E., Andrew A. Biewener, Thomas A. McMahon, Peter G. Weyand, and Hugh M. Herr. 2002. “Energetics and Mechanics of Human Running on Surfaces of Different Stiffnesses.” Journal of Applied Physiology 92 (2): 469–78. doi:10.1152/japplphysiol.01164.2000.

Kuo, Arthur D. 2002. “Energetics of Actively Powered Locomotion Using the Simplest Walking Model.” Journal of Biomechanical Engineering 124 (1): 113–20. doi:10.1115/1.1427703.

Kuo, Arthur D., J. Maxwell Donelan, and Andy Ruina. 2005. “Energetic Consequences of Walking Like an Inverted Pendulum: Step-to-Step Transitions.” Exercise & Sport Sciences Reviews 33 (2): 88–97.

Lafortune, M. A., and E. M. Hennig. 1992. “Cushioning Properties of Footwear during Walking: Accelerometer and Force Platform Measurements.” Clinical Biomechanics 7 (3): 181–84. doi:10.1016/0268-0033(92)90034-2.

Margaria, R., P. Cerretelli, P. Aghemo, and G. Sassi. 1963. “Energy Cost of Running.” Journal of Applied Physiology 18 (2): 367–70.

Martin, PE. 1985. “Mechanical and Physiological Responses to Lower Extremity Loading during Running.” Medicine and Science in Sports and Exercise 17 (4): 427–33.

McMahon, Thomas A., and George C. Cheng. 1990. “The Mechanics of Running: How Does Stiffness Couple with Speed?” Journal of Biomechanics, International Society of Biomechanics, 23, Supplement 1: 65–78. doi:10.1016/0021-9290(90)90042-2.

Mochon, Simon, and Thomas A. McMahon. 1980. “Ballistic Walking.” Journal of Biomechanics 13 (1): 49–57. doi:10.1016/0021-9290(80)90007-X.

Neptune, R. R., F. E. Zajac, and S. A. Kautz. 2004. “Muscle Mechanical Work Requirements during Normal Walking: The Energetic Cost of Raising the Body’s Center-of-Mass Is Significant.” Journal of Biomechanics 37 (6): 817–25. doi:10.1016/j.jbiomech.2003.11.001.

New Balance. 2015. “First 3D-Printed Running Shoe.” New Balance. November 24. http://www.newbalance.com/article?id=4041.

Nigg, B. M., H. A. Bahlsen, S. M. Luethi, and S. Stokes. 1987. “The Influence of Running Velocity and Midsole Hardness on External Impact Forces in Heel-Toe Running.” Journal of Biomechanics 20 (10): 951–59. doi:10.1016/0021-9290(87)90324-1.

Nigg, B. M, D Stefanyshyn, G Cole, P Stergiou, and J Miller. 2003. “The Effect of Material Characteristics of Shoe Soles on Muscle Activation and Energy Aspects during Running.” Journal of Biomechanics 36 (4): 569–75. doi:10.1016/S0021-9290(02)00428-1.

Nigg, Benno M. Dr sc nat. 1997. “Impact Forces in Running.” Current Opinion in Orthopaedics 8 (6): 43–47.

Owings, Tammy M., and Mark D. Grabiner. 2003. “Measuring Step Kinematic Variability on an Instrumented Treadmill: How Many Steps Are Enough?” Journal of Biomechanics 36 (8): 1215–18. doi:10.1016/S0021-9290(03)00108-8.

Radin, Eric L., Howard G. Parker, James W. Pugh, Robert S. Steinberg, Igor L. Paul, and Robert M. Rose. 1973. “Response of Joints to Impact Loading — III.” Journal of Biomechanics 6 (1): 51–57. doi:10.1016/0021-9290(73)90037-7.

Ruina, Andy, John E. A. Bertram, and Manoj Srinivasan. 2005. “A Collisional Model of the Energetic Cost of Support Work Qualitatively Explains Leg Sequencing in Walking and Galloping, Pseudo-Elastic Leg Behavior in Running and the Walk-to-Run Transition.” Journal of Theoretical Biology 237 (2): 170–92. doi:10.1016/j.jtbi.2005.04.004.

Sarrafian, Shahan K. 1987. “Functional Characteristics of the Foot and Plantar Aponeurosis under Tibiotalar Loading.” Foot & Ankle International 8 (1): 4–18. doi:10.1177/107110078700800103.

Sawicki, Gregory S., Cara L. Lewis, and Daniel P. Ferris. 2009. “It Pays to Have a Spring in Your Step.” Exercise and Sport Sciences Reviews 37 (3): 130. doi:10.1097/JES.0b013e31819c2df6.

Schwartz, Michael H., and Adam Rozumalski. 2005. “A New Method for Estimating Joint Parameters from Motion Data.” Journal of Biomechanics 38 (1): 107–16. doi:10.1016/j.jbiomech.2004.03.009.

Siegel, Karen Lohmann, Thomas M. Kepple, and Graham E. Caldwell. 1996. “Improved Agreement of Foot Segmental Power and Rate of Energy Change during Gait: Inclusion of Distal Power Terms and Use of Three-Dimensional Models.” Journal of Biomechanics 29 (6): 823–27. doi:10.1016/0021-9290(96)83336-7.

Skinner, Nathaniel E., Karl E. Zelik, and Arthur D. Kuo. 2015. “Subjective Valuation of Cushioning in a Human Drop Landing Task as Quantified by Trade-Offs in Mechanical Work.” Journal of Biomechanics 48 (10): 1887–92. doi:10.1016/j.jbiomech.2015.04.029.

Takahashi, Kota Z., Thomas M. Kepple, and Steven J. Stanhope. 2012. “A Unified Deformable (UD) Segment Model for Quantifying Total Power of Anatomical and Prosthetic below-Knee Structures during Stance in Gait.” Journal of Biomechanics 45 (15): 2662–67. doi:10.1016/j.jbiomech.2012.08.017.

Takahashi, Kota Z., and Steven J. Stanhope. 2013. “Mechanical Energy Profiles of the Combined Ankle–foot System in Normal Gait: Insights for Prosthetic Designs.” Gait & Posture 38 (4): 818–23. doi:10.1016/j.gaitpost.2013.04.002.

Tung, Kryztopher David, Jason R Franz, and Rodger Kram. 2014. “A Test of the Metabolic Cost of Cushioning Hypothesis during Unshod and Shod Running.” Med Sci Sports Exerc 46 (2): 324–29.

Umberger, Brian R. 2010. “Stance and Swing Phase Costs in Human Walking.” Journal of The Royal Society Interface 7 (50): 1329–40. doi:10.1098/rsif.2010.0084.

Voloshin, Arkady, and Josef Wosk. 1981. “Influence of Artificial Shock Absorbers on Human Gait.” Clinical Orthopaedics & Related Research 160 (October): 52–56.

Winter, David A., H. Grant Sidwall, and Douglas A. Hobson. 1974. “Measurement and Reduction of Noise in Kinematics of Locomotion.” Journal of Biomechanics 7 (2): 157–59. doi:10.1016/0021-9290(74)90056-6.

Yeom, Jin, and Sukyung Park. 2011. “A Gravitational Impulse Model Predicts Collision Impulse and Mechanical Work during a Step-to-Step Transition.” Journal of Biomechanics 44 (1): 59–67. doi:10.1016/j.jbiomech.2010.08.024.

Zelik, Karl E., Tzu-Wei P. Huang, Peter G. Adamczyk, and Arthur D. Kuo. 2014. “The Role of Series Ankle Elasticity in Bipedal Walking.” Journal of Theoretical Biology 346 (April): 75–85. doi:10.1016/j.jtbi.2013.12.014.

Zelik, Karl E., and Arthur D. Kuo. 2010. "Human Walking Isn't All Hard Work: Evidence of Soft Tissue Contributions to Energy Dissipation and Return." Journal of Experimental Biology 213 (24): 4257–64. doi:10.1242/jeb.044297.

Zelik, Karl E., Kota Z. Takahashi, and Gregory S. Sawicki. 2015. "Six Degree-of-Freedom Analysis of Hip, Knee, Ankle and Foot Provides Updated Understanding of Biomechanical Work during Human Walking." Journal of Experimental Biology 218 (6): 876–86. doi:10.1242/jeb.115451.

Zeni Jr., Joseph A., and Jill S. Higginson. 2010. "Gait Parameters and Stride-to-Stride Variability during Familiarization to Walking on a Split-Belt Treadmill." Clinical Biomechanics 25 (4): 383–86. doi:10.1016/j.clinbiomech.2009.11.002.

APPENDIX

Subject Consent Form

Vanderbilt University Institutional Review Board
Informed Consent Document for Research

Principal Investigator: Karl Zelik, Ph.D. Revision Date: October 21, 2014
Study Title: Effect of walking aids on locomotor performance
Institution/Hospital: Vanderbilt University (Nashville, TN)

This informed consent applies to: Adults 18 – 75

Name of participant: ____ Age: ___________

The following is given to you to tell you about this research study. Please read this form with care and ask any questions you may have about this study. Your questions will be answered. Also, you will be given a copy of this consent form.

You do not have to be in this research study. You can stop being in this study at any time. If we learn something new that may affect the risks or benefits of this study, you will be told so that you can decide whether or not you still want to be in this study.

1. What is the purpose of this study?

You are being asked to be in this research study to help develop fundamental knowledge about healthy human locomotion, which can then motivate improvements in lower-limb walking aids, such as orthoses and prostheses. Specifically, the researchers seek to perform a series of experiments on healthy participants aimed at better understanding biomechanical contributions of muscles and tendons acting at the hip, knee, ankle and foot. There are over 30 million Americans living with locomotor impairments, ranging from individuals with lower-limb amputation to incomplete spinal cord injury to congenital foot or leg disorders. The mobility-related challenges experienced by these individuals have been shown to restrict quality of life, make daily activities more difficult, reduce physical activity and social engagement, and increase the risk of developing additional health problems (e.g., back pain, osteoarthritis). Innovative assistive devices could dramatically improve the health and mobility of these individuals by augmenting human movement, but only if properly designed and integrated with the human body. In order to best utilize technological advancements to assist locomotion for disabled individuals we must first understand healthy human locomotion, specifically the biological mechanisms that enable healthy, safe and efficient movement.

2. What will happen and how long will you be in the study?

If you choose to participate, you will be asked to participate in one, or possibly a series, of experimental sessions. You will perform testing on a treadmill, and may be asked to perform various locomotor tasks (e.g., walking, jogging, skipping, hopping, walking sideways, walking up or down an incline). For some conditions you may also be asked to use a specified walking aid. Walking aids may include a specific type of shoe (which will be provided by the experimenters), or an orthotic brace such as an ankle brace or knee brace. Some walking aids may be commercially-available products, which are typically used by individuals recovering from injury or individuals with walking difficulties. Some walking aids may be custom-designed, but similar to the commercially-available products. Custom-designed walking aids will be used as a way to systematically and experimentally investigate various scientific hypotheses. It will always be your choice whether or not you want to participate. Your decision to not participate on a given day or at any time during the experiment will not affect the outcome of the study. At any point you may decide to stop being in the study altogether, in which case all you need to do is to inform the researchers.

To participate in this study, you must be a self-reported healthy adult (18-75 years old) with no known locomotor or neurophysiological disorders that would impair your gait. You will be excluded if you have any recent compromising health factors, such as a recent bone fracture or acute injury. Experimental tests will take place in the Biomechanics and Assistive Technology Laboratory in Olin Hall 506. Testing will utilize common non-invasive gait analysis techniques and instruments, such as a force-instrumented treadmill to measure the ground reaction forces exerted by the body, a motion capture system to record body movement, an electromyographic system to measure muscle activity, an ultrasound system to measure muscle fascicle length changes and a respirometry system to measure oxygen consumption. The treadmill that will be used for testing is inclinable, and some tests may involving walking up or down a fixed slope. Testing conditions may involving walking at different speeds or step cadences, and may be performed with or without shoes (for the latter condition participants will be offered disposable foot/shoe covers to use).

Experimental sessions may vary in length, but will typically last about 2-4 hours. During an experimental we may ask you to wear lycra spandex shorts and top, to enable us to accurately record your body movements. We will place some sensors on you (e.g., reflective markers for motion capture, wireless electromyography electrode for measuring muscle activity), which will be affixed via hypoallergenic tape or Velcro straps. To measure muscle activity accurately we may also need to shave small (roughly 1 inch by 1 inch) patches using a disposable razor (to ensure clean electrode contact with the skin and no interference from hair). We will ask you before doing so, and you are welcome to deny our request. During these sessions you will be given as much time as you desire to rest or simply to stop and ask questions. At any point in time you may decide to end a session for any reason, simply inform the experimenter.

3. Costs to you if you take part in this study:

There is no cost to you for taking part in this study.

4. Side effects and risks that you can expect if you take part in this study:

The risks involved in this research are similar to those of exercising on a treadmill at low to moderate speed. There is a risk of tripping or falling, but for any conditions where balance is a concern these risks are lessened by having an emergency treadmill stop button, a treadmill hand rail and/or a ceiling-mounted safety harness system.

5. Risks that are not known:

Muscle soreness or pain from the treadmill exercise is also a risk, and may depend on your current level of fitness. Because some devices are investigational, meaning non-FDA approved, there may be risks that we do not know about at this time.

6. Payment in case you are injured because of this research study:

If it is determined by Vanderbilt and the Investigator that an injury occurred as a direct result of the tests or treatments that are done for research, then you and/or your insurance will not have to pay for the cost of immediate medical care provided **at Vanderbilt** to treat the injury.

There are no plans for Vanderbilt to pay for the costs of any additional care. There are no plans for Vanderbilt to give you money for the injury.

7. Good effects that might result from this study:

a) The benefits to science and humankind that might result from this study are: The results may lead to a deeper understanding of healthy human locomotion, and to advances in assistive technologies such as foot prostheses for amputees or lower-limb orthoses for individuals with physical disabilities.
b) The benefits you might get from being in this study are: There may be no direct benefit to you for taking part in this study. The benefits you might receive are improved health associated with gait training. Some of the devices being tested are investigational and not available for purchase.

8. Other treatments you could get if you decide not to be in this study:

This study provides no therapeutic benefits.

9. Payments for your time spent taking part in this study or expenses:

You may be reimbursed for your time, at a rate of $10 per hour.

10. Reasons why the study doctor may take you out of this study:

You will be removed from the study if there is any reason for concern for your safety or if you are unable to complete the tasks for any reason. If you are removed from the study, you will be told the reason.

11. What will happen if you decide to stop being in this study?

If you decide to stop being part of the study, you should tell the experimenter. Your decision to not participate will not affect the outcome of the study.

12. Who to call for any questions or in case you are injured:

If you should have any questions related to this specific study or you feel you have been hurt by being a part of this study, you can contact the study director, Dr. Karl Zelik, Vanderbilt University, Biomechanics and Assistive Technology Laboratory at (615) 875-1506; email karl.zelik@vanderbilt.edu.

For additional information about giving consent or your rights as a person in this study, to discuss problems, concerns, and questions, or to offer input, please feel free to call the Vanderbilt University Institutional Review Board Office at (615) 322-2918 or toll free at (866) 224-8273.

13. Confidentiality and Authorization to Use/Disclose Protected Health Information

Federal regulations give you certain rights related to your Protected Health Information (PHI). These include the right to know who will be able to get the information and why. The researchers must get your authorization (permission) to use or release any health information that might identify you.

Is my health information protected after it has been given to others?

All data, video, and photographs recorded during this study will be held in the Biomechanics and Assistive Technology Laboratory at Vanderbilt University on a secure computer or hard drive. Data will continue to be held until Dr. Zelik (the study director) deems the data is no longer scientifically relevant, at which point the data will be destroyed.

All efforts, within reason, will be made to keep your personal information in your research record confidential, but total confidentiality cannot be guaranteed. The results of this participation will be anonymous and will not be released in any individually identifiable form without your prior written consent, unless required by law.

A unique study ID number will be assigned to you and will be used on all study data records. The file that links your personally identifiable information with the unique study ID number assigned to you will be kept in a securely locked file only accessible to the research staff. Your name and any other identifying information collected will be kept in a locked file drawer only accessible to research staff. These precautions are expected to be completely effective in eliminating risks to confidentiality.

What if I decide not to give permission to use and give out my health information?

If you decide not to give permission to use your health information, you will not be enrolled in the study, but there will be no negative effects on you.

May I withdraw or revoke (cancel) my permission?

You have the right to revoke this authorization and can withdraw your permission for us to use your information for this research by sending a written request to the Principal Investigator listed on page one of the research consent form. If you do send a letter to the Principal Investigator, the use and disclosure of your protected health information will stop as of the date he receives your request. However, the Principal Investigator is allowed to use and disclose information collected before the date of the letter or collected in good faith before your letter arrives.

May I review or copy the information obtained from me or created about me?

After the study is completed and the results have been analyzed, you will be permitted access to any data that was collected about you in the study.

Vanderbilt may share your information, without identifiers, to others or use it for other research projects not listed in this form. Vanderbilt, Dr. Zelik and his staff will comply with any and all laws regarding the privacy of such information. There are no plans to pay you for the use or transfer of this de-identified information.

STATEMENT BY PERSON AGREEING TO BE IN THIS STUDY

I have read this consent form and the research study has been explained to me verbally. All my questions have been answered, and I freely and voluntarily choose to take part in this study.

______________________ ______________________________________

Date Signature of patient/volunteer

Consent obtained by:

______________________ ______________________________________

Date Signature

____

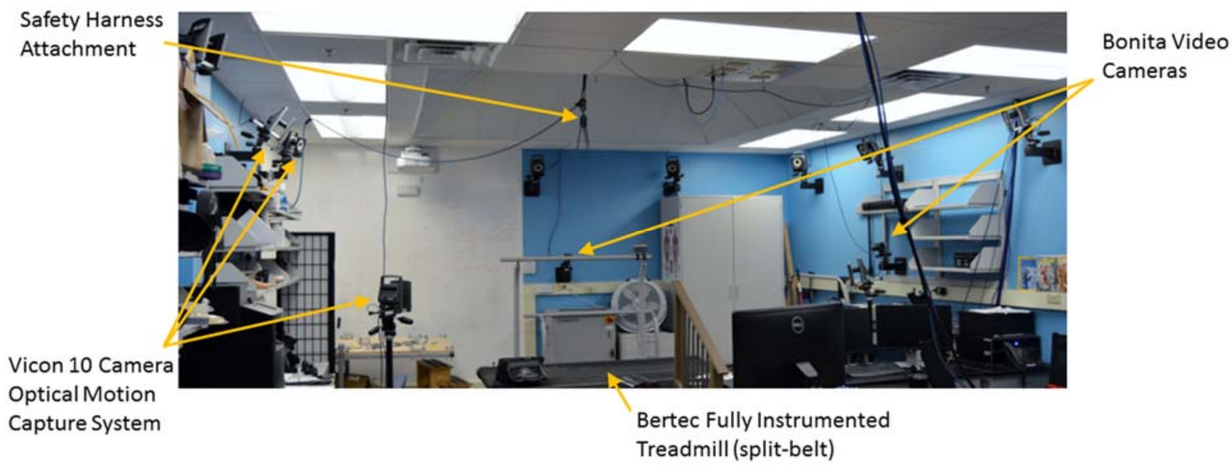
Printed Name and Title

Marker Set on Subject

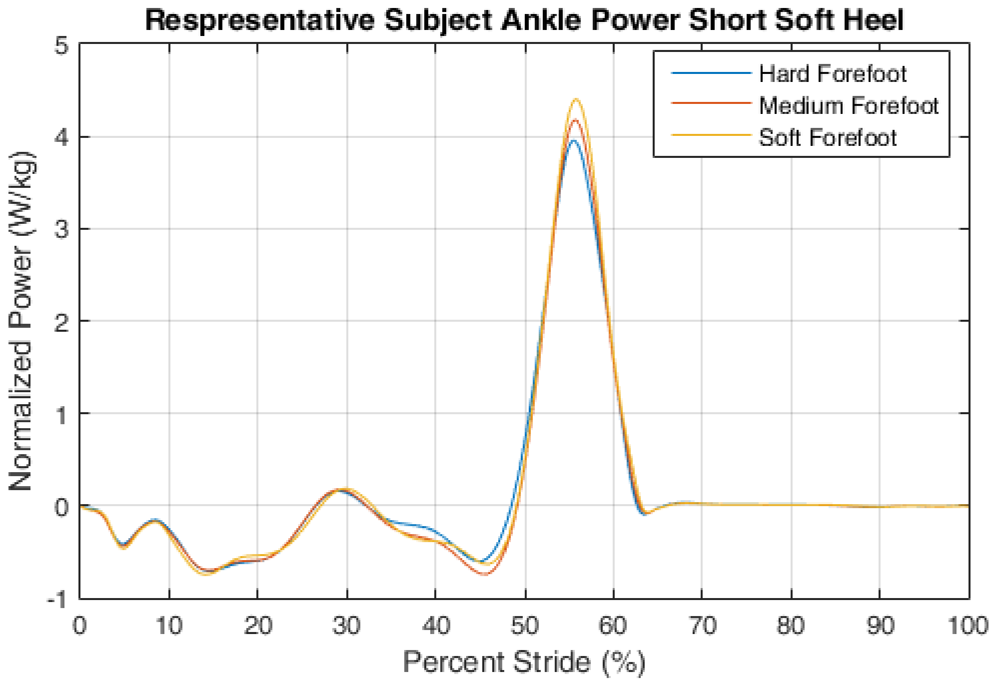
Pelvis	• Anterior Superior Illiac Spine • Posterior Superior Illiac Spine
Thigh	• Greater Trochanter • Medial Epicondyle • Lateral Epicondyle • 4 Marker Cluster
Shank	• Medial Malleolus • Lateral Malleolus • 4 Marker Cluster
Foot	• Calcaneus (approximation when shod) • 1st Metatarsal (approximation when shod) • 5th Metatarsal (approximation when shod)

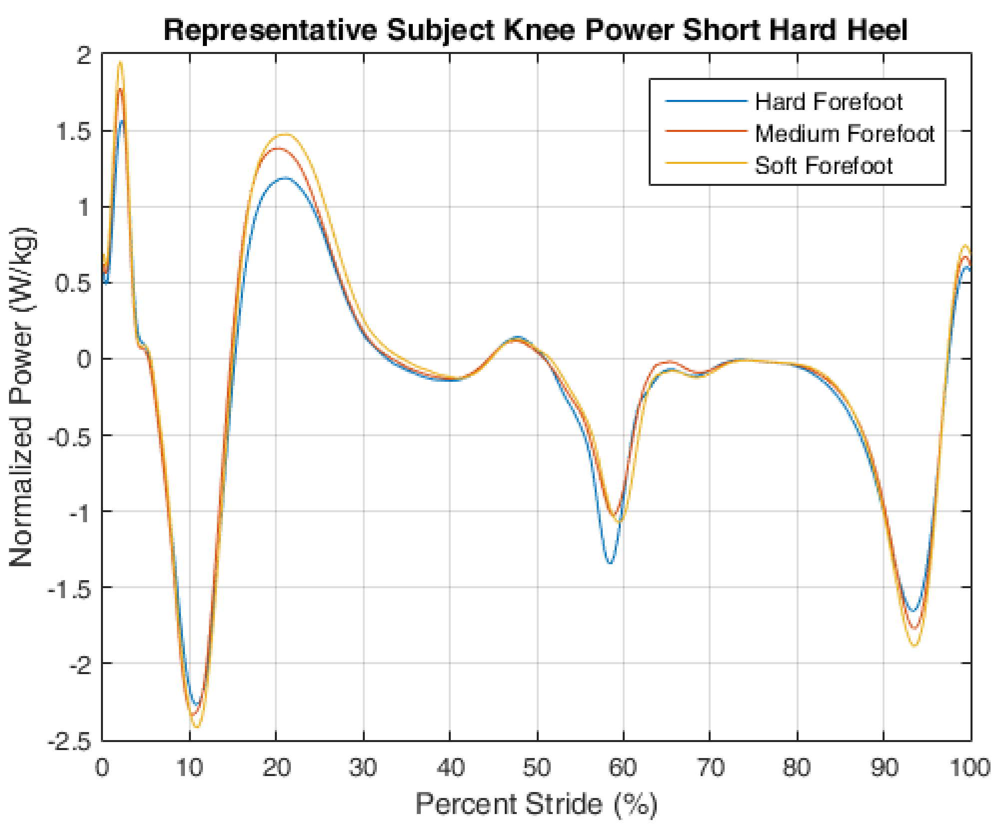
*All markers above doubled for right and left sides of body

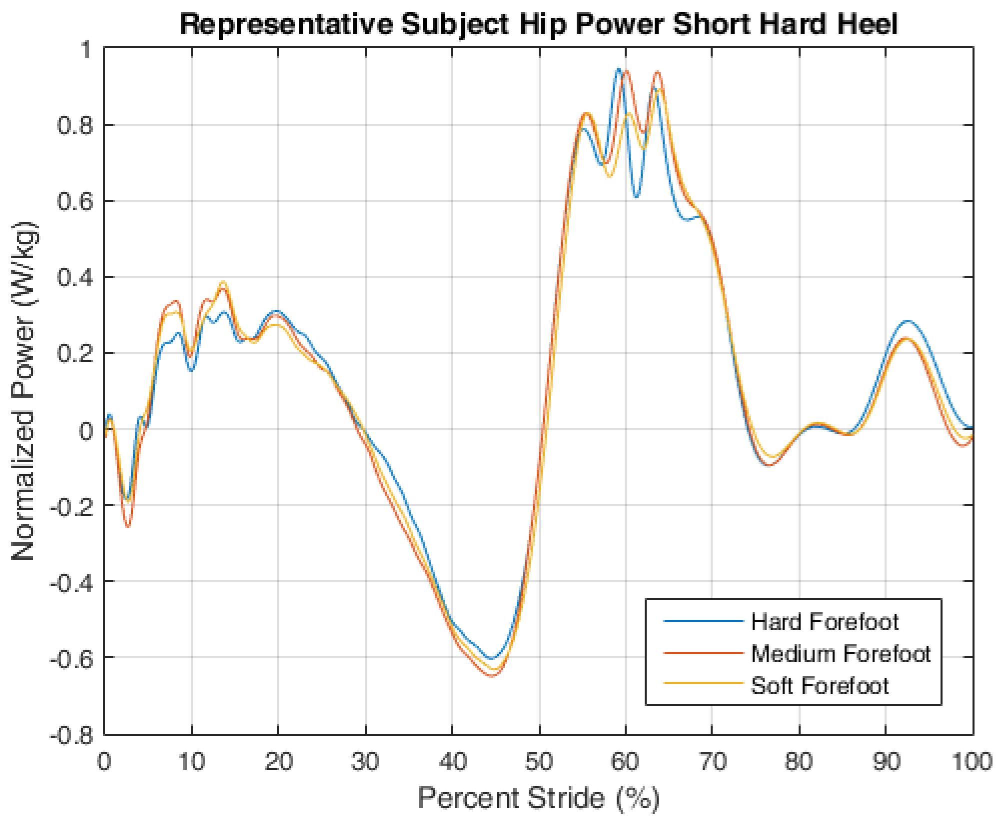
Lab Setup and Subject Mid-Experiment



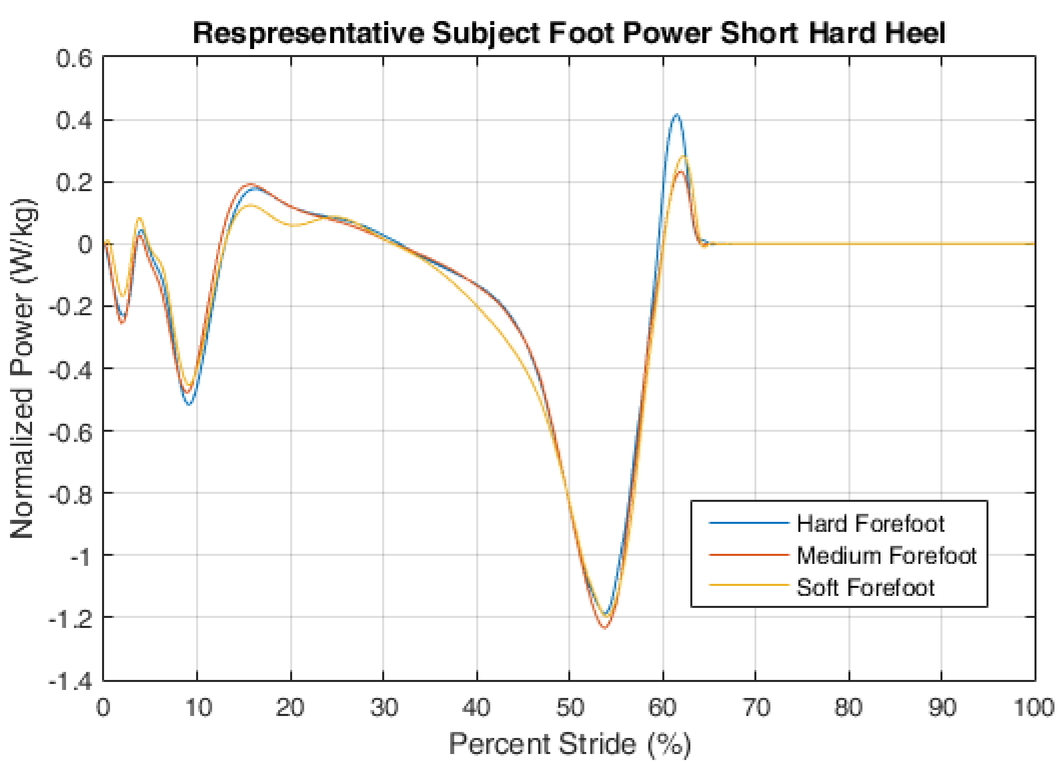
Joint and Foot Push-Off Power Trends





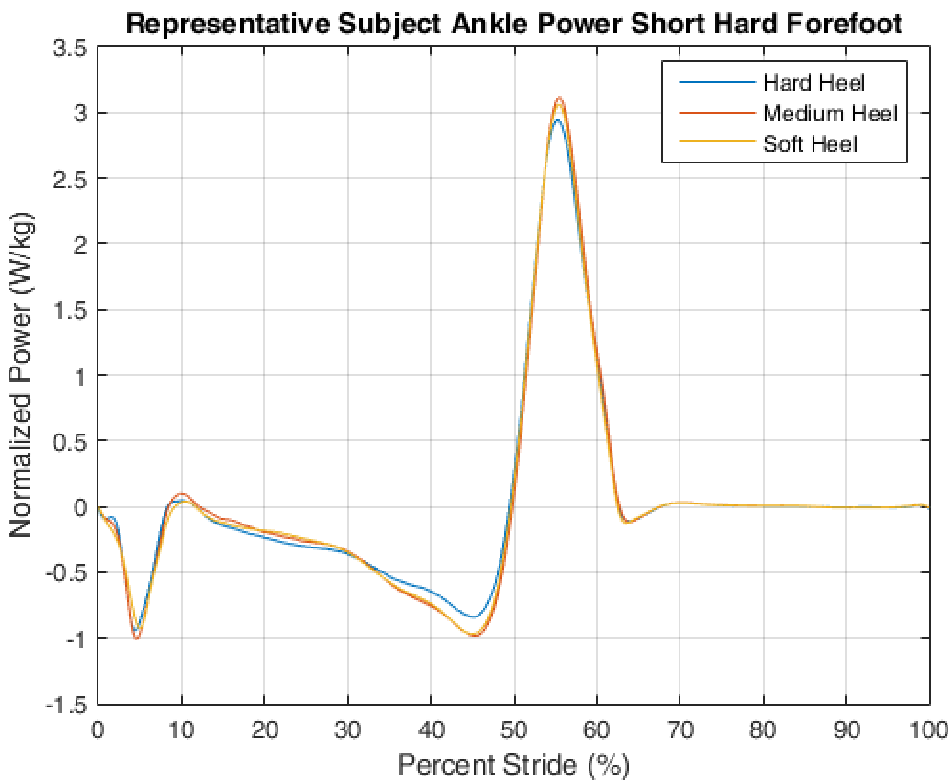


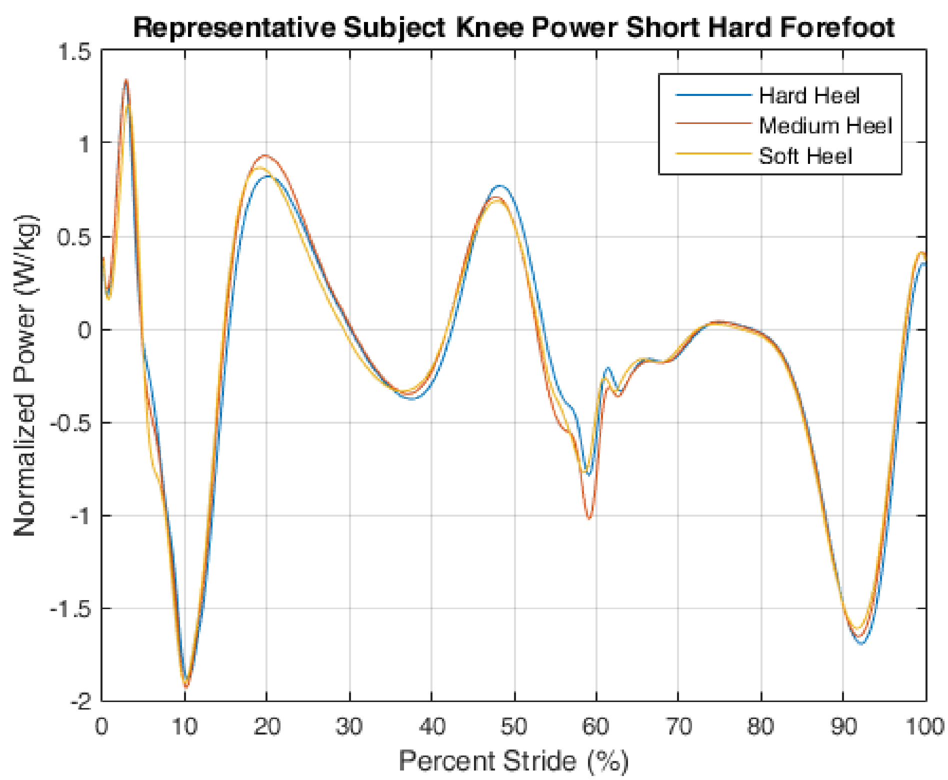
Representative Subject Hip Power Short Hard Heel
Normalized Power (W/kg)
Percent Stride (%)
Hard Forefoot
Medium Forefoot
Soft Forefoot

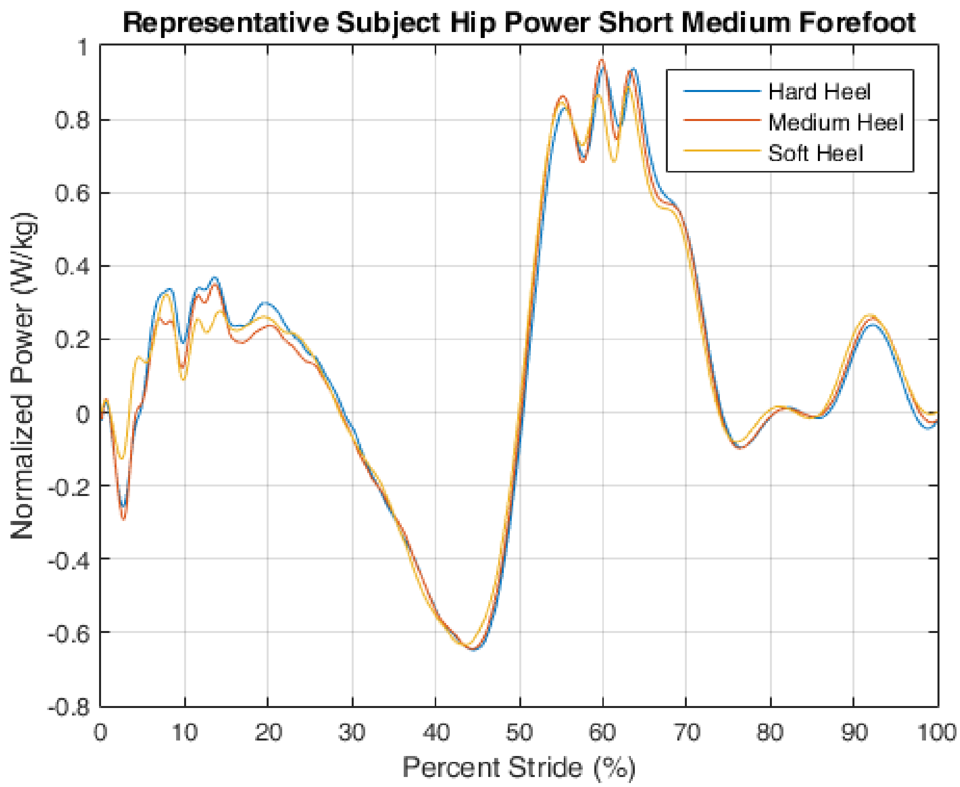


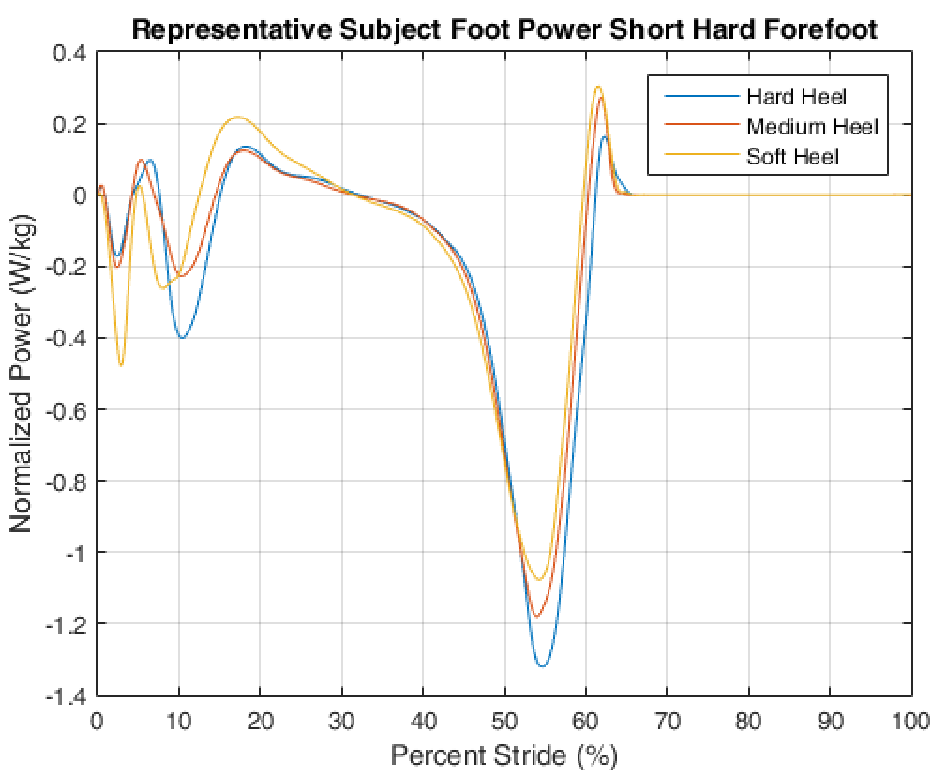
Respresentative Subject Foot Power Short Hard Heel
Normalized Power (W/kg)
Percent Stride (%)
Hard Forefoot
Medium Forefoot
Soft Forefoot

Joint and Foot Collision Power Trends






Representative Subject Hip Power Short Medium Forefoot
Hard Heel
Medium Heel
Soft Heel
Normalized Power (W/kg)
Percent Stride (%)


Representative Subject Foot Power Short Hard Forefoot
Hard Heel
Medium Heel
Soft Heel
Normalized Power (W/kg)
Percent Stride (%)

Converting from Normalized Units to Dimensionless Units (Using Subject Average Leg Length)

	Parameter One		Parameter Two		Constant	Formula
To convert	Normalized Power (W/kg)	to	Dimensionless Power	divide parameter one by	29.54 W/kg	g*sqrt(g*L)
	Normalized Work (J/kg)		Dimensionless Work		9.072 J/kg	g*L